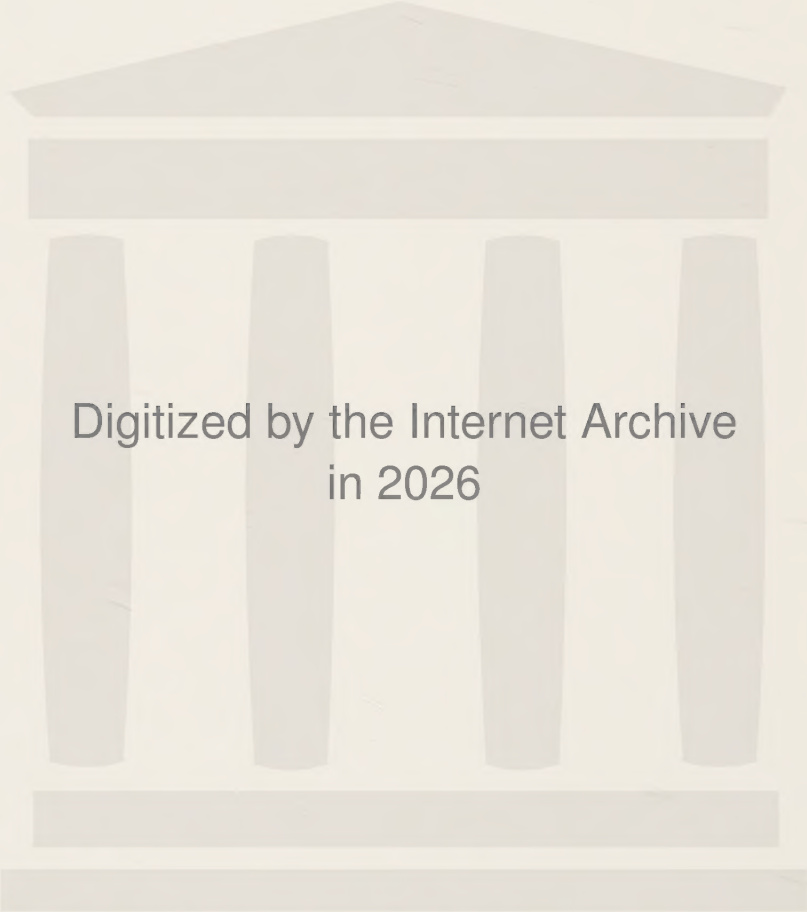


Thermal Aids in Urologic Surgery

by
S. Björn Lundquist



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THERMAL AIDS IN UROLOGIC SURGERY

Stig Björn Lundquist

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Akademisk avhandling

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Abstract The purpose of this study has been to investigate the physical and physiological events in urologic thermo-surgery, in order to facilitate the dialogue between surgeon and physicist, necessary for user impact on equipmental development. A review of historical and present use of thermal aids in urologic surgery is given. The mechanisms responsible for electrosurgical cutting are theoretically analyzed and followed by experimental and clinical investigations. The lack of objective information concerning the surgical performance of electrosurgical equipment is demonstrated, and suggestions made to what such information should contain. The development of a surgical invasive microwave antenna for transurethral in depth coagulation of bladder tumors is described. Near field antenna theories are presented. Experimental investigations of the antenna performance are reported. A theoretical model for estimation of the tissue damage in CO₂ laser surgery is presented and correlated to experimental investigations of the microvascular injury produced. The author's experience of CO₂ laser surgery in urology is presented, including a study in 90 patients of the efficiency of CO₂ laser therapy in the management of podophyllin resistant condylomata acuminata. Indications for the CO₂ laser in urologic surgery are suggested and the present use of other lasers in urology is reviewed.		
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CONTENTS

INTRODUCTION 4.

INTERACTION OF THERMAL ENERGY WITH BIOLOGICAL
TISSUE 5.

ACTUAL CAUTERY 11.

ELECTROSURGERY 12.

DEVELOPMENT OF AN INVASIVE MICROWAVE
HEATING METHOD FOR TRANSURETHRAL IN DEPTH COAGULA-
TION OF TUMORS. 40

LASERS 64.

SUMMARY 74.

ACKNOWLEDGEMENTS 75.

LITERATURE 76.

TABLES 85.

Introduction

Thermal energy has since antiquity been an important tool in surgery and medicine. Mild heat has been used to treat a variety of conditions. The initial sources of mild heat were found in nature, e.g. the sun, warm sand and thermal baths. The use of intense heat for hemostasis was described several thousand years ago. The original sources of such heat were heated oils or metals and burning-glasses.

The technological development during this century has made a number of modalities for mild and intense heating available to us. The control of heating is improving, and the technology involved tends to expand. The mechanisms of interaction with biological tissue are complicated and the resulting surgical effects become increasingly difficult to understand for users of thermo-surgical equipment. As users we ought to have an impact on the development of our instruments. This is, however, difficult to realize without a better knowledge of the basic physics. The purpose of the present investigation has been to analyze the problems of urological surgeons in terms of physics and present the relevant physical and physiological facts in a manner perceivable to medical as well as technical specialists.

Interaction of thermal energy with biological tissue

Present use of thermal energy in surgery

The use of moderate heating (42–45°C) for hyperthermia in cancer therapy is becoming well-established. Though this application is not conforming to the traditional concept of surgical heat, the introduction of the heating device may require surgical intervention.

More intense heating (above 60°C) has well defined applications, which are:

- 1) Hemostasis or cutting with simultaneous hemostasis in open surgery.
- 2) Cutting, tissue removal and hemostasis in endoscopic procedures.
- 3) Destruction of tumors or other diseased tissue in open or endoscopic surgery.

Temperature and tissue effects

Surgical heat may be defined as temperatures where irreversible tissue damage occurs within seconds. The survival rate of mammalian cells exposed to elevated temperatures for different time can be estimated from Arrhenius' integral plots. Normal cells are considered to survive prolonged exposure to temperatures below 44°C, while tumor cells are more susceptible. This forms the basis of hyperthermia treatment. Between 58 and 63°C, death of normal cells occurs within a few seconds. Above 63°C it is instantaneous. Above 58°C the mechanical and structural tissue properties are altered. The first obvious change is denaturing of proteins. Above 70°C collagenic structures are denatured which causes contraction of tissue and decreased stability. Closer to 100°C loss of tissue water occurs, being complete at 100°C. The nature of the changes in the 100–200°C range is incompletely known. Approaching 250°C carbonization begins, and at 350–400°C disintegration of tissue occurs.

Heat transfer to tissue

Heating of tissue can be effected by different kinds of external energy sources. Except for the case of a primarily thermal energy source, the primary energy is converted into thermal energy within the tissue, due to the exertion of a kind of mechanical work. The external source delivers a certain amount of energy per unit time – the power. The power density is the power per unit area or volume. This thermal power density induced in the tissue by the external source gives rise to a certain thermal flux density. The units of measurement and the symbols are:

Energy W , Joules (J); Power P , Watts (W); Power density q , Wm^{-3} (or Wm^{-2}); and Thermal flux density Q , Wm^{-2} .

Temperature represents the state of energy of a medium. The energy absorbed by a certain tissue volume thus could be expected to determine the resulting temperature. However, energy is lost from the heated volume into the surrounding medium. This process takes time. In surgery the resulting temperature therefore is determined by the absorbed energy density and the time during which it is delivered.

Distribution of heat in tissue

When a tissue volume is heated, thermal flux penetrates into surrounding regions, where the externally induced power density is zero. Thus, temperature increases in the vicinity of the volume heated by external induction, while energy is lost from this region. This phenomenon depends on several mechanisms, which are influenced by tissue properties. The following terms are used to describe the thermal properties of a medium: Thermal capacitivity – C ($\text{J kg}^{-1} \text{K}^{-1}$); thermal conductivity – λ ($\text{Wm}^{-1} \text{K}^{-1}$) and thermal diffusivity –

$$D = \frac{\lambda}{\rho \cdot c} (\text{m}^2 \text{s}^{-1}) \text{ with } \rho = \text{density } (\text{kg m}^{-3}).$$

The thermal capacitivity describes the energy content per unit mass of tissue at a given temperature. This can also be expressed as the capability of the medium to store thermal energy. Consequently, this property determines the resulting temperature at a given absorbed energy density. The thermal conductivity is the proportionality constant determining the heat flux occurring at a given temperature gradient. For a given steady state thermal flux, the thermal conductivity determines the temperature gradient.

The thermal diffusivity describes the ability of a thermally perturbed medium to relax back to steady state conditions. This is the most important term for transient temperature changes within the small tissue volumes present in thermal surgery.

From a tissue heat source region energy is lost into the surrounding medium by the following mechanisms

1. Conduction is more important for long heating time, e.g. in hyperthermia, than in surgical applications. During the relaxation of the source region, heat transfer to the surroundings will cause a transient thermal perturbation in these regions. The result can be described as a heat pulse, with a decaying maximum temperature, penetrating into the surrounding. This means that tissue damage may be produced outside the source region, the extent of which depends on the time during which the thermal perturbation of the source region is maintained, and on the maximum temperature of this region.

2. Thermal convection takes place when a gas or a liquid moves in relation to a medium with a different temperature. The motion can be caused by an external pressure difference or by density variations due to varying temperature in the moving medium. The heat convection is determined by the temperature gradient and the heat transfer coefficient for the interface of the two media. It is expressed by

$$Q_{cv} = H(T_s - T_a), \text{ where}$$

$$Q_{cv} = \text{thermal flux density}$$

$$H = \text{heat transfer coefficient } (\text{Wm}^{-2} \text{K}^{-1})$$

$$T_s = \text{temperature of the tissue surface } (\text{K})$$

$$T_a = \text{temperature of the moving medium } (\text{K}).$$

In hyperthermia, convection losses to blood circulating through the area are important. In open surgery losses occur from the tissue surface to the surrounding air. $H_{\text{tissue/air}}$ is about 15. In transurethral surgery the losses are significantly larger, since $H_{\text{tissue/water}}$ is of the order of 1500.

3. Thermal radiation becomes important at extremely high temperatures. It is for a black body determined by the Stephan-Boltzmann law:

$$Q_r = s_s \sigma T_s^4, \text{ where}$$

$$\sigma = \text{the Stephan Boltzmann constant.}$$

Mathematical solution for at thermal pulse energy source

An exact description of thermal events in biological tissue is difficult because of the inhomogeneity of tissue. The action of thermal aids under different clinical conditions may, however, be estimated by using the concepts described in the previous sections. In most thermo-surgical applications external power is supplied for a short time. The heated volume is usually small and superficially located. The equation of thermal conduction in one dimension is

$$\frac{\partial T}{\partial t} = D \frac{\partial^2 T}{\partial x^2} + q/q \cdot c, \text{ where}$$

q = the source term

T = the temperature

D = the thermal diffusivity

x = the distance from the source

t = the time.

This equation is strictly applicable to a thermally insulated bar of a medium. If the surgical conditions are considered to be those of a plate along the tissue surface, as in Fig. 1, the equation can also be applied. If conditions differ from this model, an error will result in the direction of a lower temperature distributed to a larger volume than estimated from the analysis.

The free surface, $x = 0$, is effectively cooled and has a temperature assumed to be zero.

The solution for a pulsed thermal energy source concentrated at $x = 0$ at $t = 0$ can be found in textbooks of theoretical physics [1]:

$$T = \frac{e^{-x^2/4Dt}}{\sqrt{4\pi Dt}}$$

which is graphically presented in Figs. 2–3.

If T is studied as a function of t for a certain depth x , a maximum, T_1 , will be obtained at the time t . This means that

$$\frac{dT}{dt} = 0 \text{ after a certain time } t_1, \text{ which depends on } x_1.$$

Differentiation gives

$$\frac{dT}{dt} = \frac{e^{-x_1^2/4Dt_1}}{\sqrt{4\pi Dt_1}} \left[-\frac{1}{2t_1} + \frac{x_1^2}{4Dt_1^2} \right] = 0$$

which gives the following relation between x_1 and t_1

$$x_1^2 = 2 Dt_1$$

Defining the motion of the temperature front as the propagation of the maximum temperature $T_{\max}$ we find

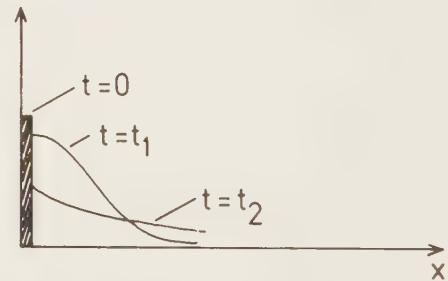
$$T_{\max} = \frac{e^{-x_1^2/4Dt_1}}{\sqrt{4\pi Dt_1}} = \frac{e^{-1/2}}{\sqrt{2\pi}} \cdot \frac{1}{x}$$

The depth of penetration of the temperature front is proportional to the square root of the time, while its amplitude is proportional to the reciprocal of the distance to the source. With increasing temperature in the source region, the temperature front moves further into the tissue before its amplitude has decayed to a value at which no tissue damage is produced. The time during which external energy is supplied to the source region affects the "width" of the moving front.

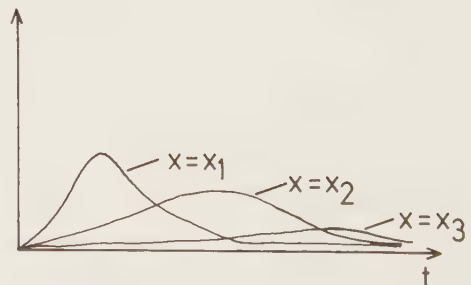
We have already discussed the temperatures necessary for producing different biological and mechanical effects in tissue. Since most thermal surgery is performed at a tissue surface it is obvious that a homogenous temperature in a larger volume of tissue cannot be produced. The primary energy sources are characterized by their heating depth, which affects the temperature gradient within the source region during the externally induced heating. The characteristics of the primary energy sources partly set the limits of clinical application, e.g. cutting and/or coagulating. However, by varying the power density and the application time or by using continuous or pulsed power, the limits of application can be extended.



1. Conditions for applying the equation of thermal conduction in one dimension.



2. Temperature as a function of distance to a thermal pulse energy source at different times.



3. Temperature at different distances to a thermal pulse energy source as a function of time.

Temperature and surgical effect

1. Thermal destruction of tumors or other diseased tissue.

a) The tissue volume is heated to at least 60°C. This temperature is sufficient to produce a necrosis, which is disposed of by sloughing and/or resorption.

b) The tissue is removed by heating to about 400°C, whereby disintegration occurs.

2. Hemostasis

a) In open surgery, the bleeding vessel can usually be heated without affecting the surrounding tissue. In this case the higher temperature limit is not critical. The target volume for the primary energy is small.

b) In endoscopy the cut end of a vessel is usually level to a tissue surface. A small vessel with low pressure may be sealed by superficial charring. In larger vessels this is not possible. Investigations have shown that heating causes contraction of the vessel lumen through denaturing of collagen fibrils in the wall [2, 3]. This process begins at 70°C and has reached its maximum at 90°C. Only a few seconds are required for the shrinkage. Arteries have to be heated to higher temperatures than veins to reach maximum contraction. Heating by 90°C did not produce permanent occlusion of the vessels. The reduced vessel lumen presents a significant flow resistance, which also depends on the length of the contraction. The open surgical vessel occlusion above 250°C occurs because the vessel is carbonized to an amorphous sealing plug. In transurethral hemostasis we will have to rely on heating the vessel to 75–90°C, for the maximum possible length, to reduce the pressure. This has to be combined with the production of a superficial carbonization, or thrombus. The in-depth heating may be difficult with heating modalities available in transurethral procedures, and always includes a damage to the surrounding tissue. With modalities involving mechanical tissue contact even the carbonization may prove difficult because the instrument tends to stick to the tissue. At the removal of the instrument bleeding may start again, and it will not be easier to stop at the second attempt.

If it is possible to maintain a temperature at 100°C, the loss of tissue water may produce an additional reduction of the vessel lumen. It is well-known that occlusion in moderate size arteries is possible by heating to 95–100°C for a few seconds [4].

3. Thermal cutting can be described as disintegration of tissue along a narrow line. The tissue surface temperature is about 400°C. The thermal load per unit area of tissue depends on the cutting velocity. To make the incision line narrow, the source must allow focusing of energy. If the primary energy penetrates into the tissue, it must not induce a thermal tissue source region of large depth. The depth of thermal damage also depends on the applied power density. The intense heat at the surface occludes small vessels as they are cut.

Considerations on the depth of tissue damage at high surface temperatures.

A great deal of the thermal energy will be lost into the external medium. The energy dissipated within the tissue will, to a major part, be consumed by the evaporation of water. It can therefore be assumed that the maximum temperature causing thermal diffusion into the tissue is 100°C, and the temperature decays to a level which does not produce tissue damage, within a short dis-

tance. In cutting, the thermal perturbation of each tissue portion is transient. If tissue is removed over a large area by the same principle, the time of thermal perturbation is longer, which gives rise to a somewhat deeper tissue damage. An important implication is that the in-depth tissue damage with a non-penetrating energy source is less sensitive to prolonged exposure of individual tissue portions than it is with a penetrating energy source.

Additional literature [5–7]

Some technical modalities for the transfer of heat to tissue

The following thermal aids will be discussed in this text:

- 1) Actual cautery
- 2) High frequency alternating currents – electrosurgery
- 3) High frequency electromagnetic fields – microwaves
- 4) Lasers

Some modalities reported by other investigators will not be discussed here:

- 5) The plasma scalpel [8]
- 6) The ultrasonic scalpel [9, 10]

Actual cautery

The word cautery originates in the Greek *καυτήριον*, which means branding iron. As indicated, the primary energy source is thermal. Cauterization, however, also means destruction of tissue by chemical substances. The original use of thermal and chemical cautery in surgery resulted in a purulent secretion, which was considered beneficial in the treatment of wounds. Infected wounds presented a major problem in military surgery. When fire-arms became common, these problems increased. Gun-shot wounds were considered to be intoxicated by gun powder, which was thought to be the explanation of their poor prognosis. It was thought, that the best way to remove these toxins, was to produce a strong suppuration. This was performed by burning the wound with branding irons or, which was later popularized by Giovanni Vigo (1460–1520), by pouring heated oil into the wounds. In 1536, Ambroise Paré was with the army of King Francis at Piémont, and the oil supply was emptied. Paré found, to his surprise, that recovery was better when no oil had been used to treat the wounds. Paré also introduced the ligature, in favor of massive cauterization, for hemostasis in amputations. He was thus a pioneer, in recognizing the importance to avoid excessive thermal tissue damage [11]. Chemical cautery was often applied transurethrally during the 19th century. The first transurethral applications of thermal cautery were made by the Italian urologist Enrico Bottini in about 1865. He used a direct current heated metal plate to treat patients with bladder neck obstruction. The results of this method were disastrous, rather than successful. A few years later, he used a similarly heated platinum wire to cut through the prostate or bladder neck. In 1874, he reported success with the method, which caused far less bleeding than other contemporary transurethral surgery [12].

At the end of the 19th century the “Bottini operation” had become popular in Europe, and remained so until replaced by electroresection.

Wires heated by direct current or by flames were used for cutting and hemostasis in open surgery. The technique is inexpensive and simple. It has some advantages for pinpoint hemostasis in delicate structures, such as the eye. Such devices with a wire tip temperature of 400–650°C are still used in ophthalmic surgery.

Electrosurgery

Several terms are used for this modality. None of them defines the mechanism of action, which is heating of tissue by a high frequency current. Some commonly used names are: surgical diathermy, high frequency (HF)-surgery, radiofrequency (RF)-surgery and electrocautery.

The development of electrosurgery

The development of surgical applications of high frequency currents is subject to a certain confusion. Even reviews made already in 1930 contain contradictory facts.

In the early development there appears to have been no linkage to surgery. The devised methods up to about 1905 all aimed at producing mild heating of larger parts of the human body.

The Yugoslavian-born physicist Nikola Tesla (1856–1943) investigated about 1890 (by then an American citizen) the biological effects of high-frequency currents. He made the important discovery that these currents did not have the hazardous effects of low frequency currents. One of his major contributions was the invention and construction of efficient generators and powerful transformers. He built a variety of devices for medical use [13, 14]. These activities, however, were a small part of his enormous production. Precise investigations were performed by the French physiologist Jacques A. d'Arsonval (1851–1940), who in 1891 reported that currents of frequencies above 10 kHz could be passed through living tissue, without causing other effects than heating [15]. The technical equipment he used was probably conforming to that designed by Tesla. It usually consisted of a direct current source with a rapidly working circuit interruptor, in combination with an inductive coil and a capacitor. Other investigators, as e.g. the American neurologist William J. Morton (1846–1920) used a simple static electricity generator connected to the patient to produce mild heating by means of moderate currents. The first to employ this knowledge clinically was the French roentgenologist Paul Oudin (1851–1923). His contribution to Tesla's equipment was a matching output inductance, the "resonator". This work was reported in 1893 [16].

The development of surgical applications seems to have been initiated by the accidental discovery that a small electrode connected to the patient could cause superficial burns and sparking. This seems to have been discovered by Joseph Riviere. The technique was improved about 1906 by the French gynecologist Samuel J. Pozzi (1846–1918). At the same time the French physician Walter Keating Hart (1870–1922) took the method into clinical use. There is some dispute whether the term fulguration (Lat. fulgur = lightning) was introduced by Pozzi or by Keating Hart. I have not been able to find an unequivocal answer in the available sources. The treatment was applied to superficial tumors by striking sparks to them. The American urologist Erwin Beer made attempts to produce underwater sparking (to skin warts) by special cystoscopic arrangements. Though this proved impossible, the warts disappeared. In 1910 he reported success in cystoscopic treatment of bladder tumors [17].

These applications were performed with a "uniterminal" connection of the patient through a single electrode. The current returned to the generator by

random pathways. The introduction of the “biterminal” application is ascribed to Doyen, who used a large metal plate as second electrode. This arrangement made a more intense current available at the active, small electrode. Doyen also used the technique for in-depth heating of large tumors. The available generators worked with an oscillating circuit and a spark gap [18].

De Forrest invented the vacuum tube in 1906 and two years later a circuit for producing a cutting current. Due to the lack of medical cooperation, this discovery was not utilized until several years later. It was subsequently found that vacuum tube generators produced smooth, clean cutting, while the spark gap generators were more suitable for coagulation and hemostasis. Later, it was made possible to coagulate with the tube generator and to cut with the gap generator. The most obvious electrical difference between the two generators was that the tube generator produced a continuous current, while the spark gap current consisted of repeated, high peak voltage, damped bursts. Since transurethral surgery put the highest demands on generator performance, urologists had an important impact on the development. In 1930 equipment combining the two generator types was mainly used. Nearly identical machines are still manufactured, and preferred by many urologists. Before this stage of development was reached, urologists with tube generators had to fight poor hemostasis, while those with spark gap generators faced the problem of poor cutting performance.

Additional literature [19–30]

Present modes of electrosurgical application and problems associated with modern electrosurgical equipment

The uniterminal connection is only used when small currents are needed, as e.g. in the fulguration of skin warts. All other applications employ the biterminal connection. The safety aspects of surgical diathermy are not discussed in this paper. Several excellent reviews exist [31–35]. The biterminal use is divided into monopolar and bipolar modes of applications. In the monopolar mode (by some termed bipolar!) one connection of the patient is made by a large ($\geq 100 \text{ cm}^2$), usually metallic, plate. This is termed “patient plate”, “neutral plate”, “neutral electrode”, “return electrode”, “inactive electrode” etc. I will refer to it as the neutral. The other connection is made by the surgical instrument, the “active”, “hot”, “live” electrode, which I refer to as the active electrode. The entire diathermy current thus passes through the patient. The small active electrode gives rise to a high current density, which is responsible for the surgical action. The large neutral (active to neutral area ratio 1:100–1:1000) produces a current density, so small that practically no heating results.

In the bipolar mode two electrodes of about the same size are used. The most common application is bipolar coagulation for hemostasis. This is usually performed with a forceps-like instrument with branches electrically insulated from one another. The current is passed mainly through the tissue grasped by the instrument. With many units a significant leakage current to earth passes through the patient. Such units may produce excessive tissue damage in delicate structures, such as the nervous system [36, 37].

Coaxial bipolar electrode arrangements are sometimes used in ophtalmic surgery. In gastro-intestinal polypectomy bipolar wire snares are sometimes used. Whenever possible, the bipolar mode should be used, since with a good electrosurgical unit, better control of tissue damage is provided, than in the monopolar mode.

Additional literature [38].

About 1960 solid state technology entered the stage. Solid state electrosurgical units (ESU) were available about 1970. In open surgery, advantages such as increased electrical safety, smaller size of equipment and decreasing demands on maintenance, have prevailed. In transurethral surgery some of the problems of the 1920s have been resurrected. Marked differences in surgical performance exist between units. These differences have not been possible to classify objectively. One of the early problems was that the semiconductors were sensitive to the temperature increase occurring in normal operation. This resulted in decreasing power and vanishing surgical performance. It was not by chance that this was first recognized in transurethral resection (TUR), since this procedure employs a high duty cycle compared to open surgery. This leaves short time for the equipment to cool between activation periods. The problem is overcome in most recently made machines. Another problem in TUR is uneven or unexpected cutting with some units, and poor hemostasis, with cutting tendencies in others. All modifications made to overcome these problems have unfortunately not been improvements. These extremes in surgical performance cannot be predicted by any of the data presented by the manufacturers. Usually the performance is described in quasi-scientific or directly non-scientific terms dating from 1930 or earlier. Technical data, except from those concerning electrical safety, often seem to have been randomly chosen and presented. Until now, it has been impossible to find two manufacturers presenting the same combination of technical specifications.

I think that modern sales organization is partly responsible for this. The early development of electrosurgery took place with a close co-operation between user and manufacturer, even in the operating room. In our time there is no such direct line, and feed-back of negative or positive experience is attenuated. Also, reports on the problems with solid state surgical diathermy are surprisingly few. An overview of literature on electrosurgery produced during 70 years, showed that practically no basic research concerning the physical and physiological mechanisms has been performed during the past 40 years. Quite a few technical innovations have been made. Those concerning electrical safety have a scientific rationale, while those meant to improve surgical performance are based upon non-scientific concepts. When these concepts are used together with electrical engineering terms, they appear strictly scientific to most users of electrosurgery.

No real development meaning improved surgical performance is possible without the cooperation of the surgeon. He is able to explain the problem in his professional terms. The electrical engineer's understanding of these terms is based on concepts, dated from the childhood of high frequency engineering, usually ill-defined and irrelevant to modern electronics. To be able to present the surgical problem in electrical engineering language I found it necessary to learn the basic facts of electricity and electronics relevant to electrosurgery.

In this work it has not been my intention to solve the technical problems. In applying surgical thinking to electricity and electric thinking to surgery, my intention has been to produce a new type of analysis of the problems as well as

a fusion of the two languages into a common, electrosurgical one. The basic concepts developed in the theoretical analysis have been tested in a series of experiments in the laboratory as well as during operations. The relevant experimental results could be interpreted and explained in terms of the theoretical concepts. Therefore I believe that my results will contribute to the future development of electrosurgery through joint work by surgeons and electrical engineers.

Traditional concepts in electrosurgery

In the beginning of electrosurgery the used terms had well-defined meanings. The different current sources gave name to currents, to which were ascribed different surgical properties (d'Arsonval c., Tesla c., Oudin c. etc.) Subsequently the surgical effects were given their own names, such as fulguration, desiccation and coagulation. A cutting current was one capable of producing cutting, etc. In more recent literature these old terms no longer have defined meanings, which is largely due to attempts to explain the connections between used current, traditional concept of the term, and tissue effect.

Fulguration still means the use of long sparks to produce wide-spread, but superficial, tissue damage (charring). The purpose may be either destruction of diseased tissue or hemostasis. In the latter meaning the synonymous term *spray coagulation* is sometimes used. It is also understood that the active electrode is not in mechanical contact with tissue in this application.

Desiccation usually means destruction of tissue without sparking by using an electrode in mechanical tissue contact. The tissue damage is said to occur by loss of cell water below boiling temperature. The electrode may be of the needle, ball or button type.

Coagulation generally is equal to hemostasis. This can be performed by passing the current through a vessel grasped by a forceps, the application being monopolar or bipolar. It is usually understood that the applied temperature is higher than in desiccation.

Pinpoint coagulation is the term used to describe hemostasis by a small electrode in mechanical tissue contact, and usually refers to endoscopic procedures. If sparking occurs during coagulation this is termed fulguration.

Cutting is explained by the rapid evaporation of cell water, which causes the cells to explode. This has been considered reinforced by histologic examinations, usually performed immediately after cutting, demonstrating very superficial changes. Sparking and arcing are considered necessary for cutting. Since different generators were used for cutting and other applications, the resulting tissue effects have been ascribed certain properties of the currents produced by these generators. Thus, a *cutting current* still means one resembling that of the electronic tube generator, that is a continuous sinusoidal waveform. A *coagulating current* consists of damped oscillations, or an otherwise modulated current. With increasing degree of modulation the coagulating effect is considered to increase. The difference in surgical action between the two generator types has been entirely ascribed to the different waveforms. The "scientific" explanation of this is that the average power, which with a continuous waveform is high, is reduced by modulation of the current. This also allows a higher peak power (and voltage) which is believed, by unexplained mechanisms, to produce a deeper heating. It is also easier to draw long sparks with the high peak voltage.

The absence of cutting action with a modulated current is explained by the low average power. At the same time it is necessary to explain why cutting does not occur when sparking is facilitated. With the coagulating current the long sparks, produced by the high peak voltage, are said to hit the tissue randomly, while in cutting the voltage is so low that the sparks are confined to a narrow line, along which the electrode is moved. The exploding cells leave microscopic holes in the tissue, which thereby is severed. When the spark gap generator is used for cutting, the breakdown frequency of the gap is doubled by adding a capacitor in the circuit. The peak voltage remains the same as for coagulating. In this case the cutting has to be ascribed to the increase in average power. This is, however, not doubled, since the energy stored by the capacitors is reduced by the addition of capacitance. Sometimes a *blended cutting current* is used. The purpose is to achieve better hemostasis during the cut. Usually the waveform of this current is a continuous sinusoidal with superimposed short bursts of higher peak amplitude.

In transurethral surgery, the hemostatic performance of the “Bovie” spark gap current is a concept, often used as a standard for comparing the performance of modern equipment. This concept is also used, when the correct term *crest factor* is used to describe hemostasis or coagulation performance. This is the ratio of peak voltage to average, or rms, voltage, when the generator is operated at no load. The higher the crest factor, the better the coagulation performance.

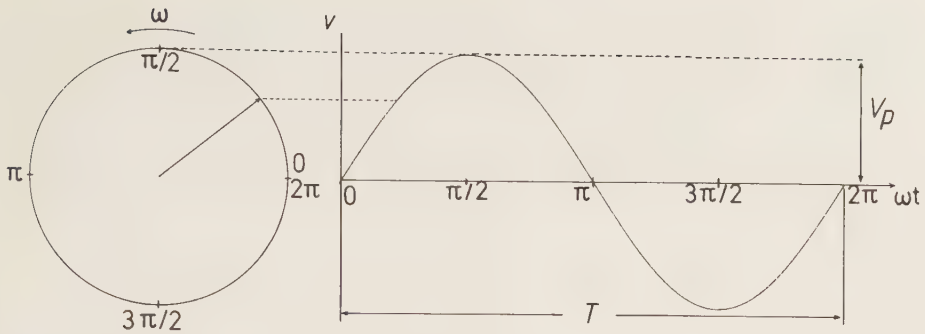
Additional literature [33, 39–45]

Some basic concepts concerning alternating currents and alternating current circuits

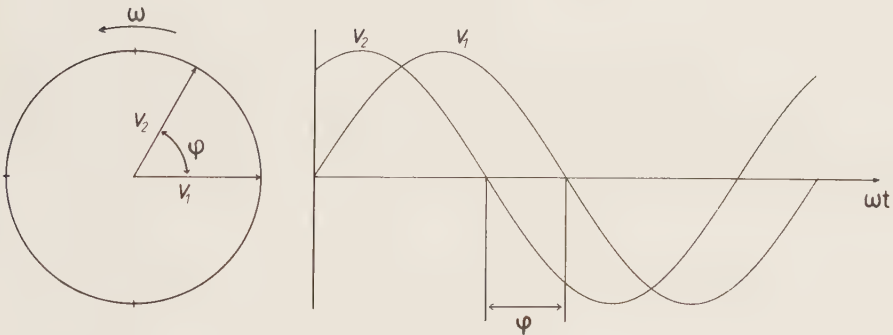
I will recapitulate some basic facts of electrical engineering and electricity as a necessary background for following discussions.

Alternating current, ac, may be generated by a variety of techniques, and have different waveforms. The simplest waveform is the sine-wave, where current and voltage vary sinusoidally with time. The sinusoidal waveform is mathematically described by the variation of the vertical component of a vector rotating counterclockwise, with a uniform angular velocity ω (Fig. 4). One complete revolution is termed a cycle and the time interval required for one cycle is called the period T . The number of cycles per second is the frequency $f = 1/T$. One complete revolution is 2π radians and requires T sec, hence

$$\omega = \frac{2\pi}{T} = 2\pi f.$$



4. Generation of sine wave. Mathematical model.



5. Two sine waves of the same frequency. Phase angle.

If the length of the vector is V_p , the instantaneous value at any time t is $v = V_p \sin \omega t$. The value V_p is the peak amplitude of the sine wave. Lower case symbols are used to indicate time-varying values, while capital letters refer to constant values or dc quantities.

A sine wave signal is completely described by its frequency, amplitude and phase angle ϕ : $v = V_p \sin (\omega t + \phi)$. Through these parameters, signals of the same frequency can be compared (Fig. 5).

To simplify the analysis of ac circuits by formally applying Ohm's law, the sinusoidal current is "translated" into a representative dc current. The effective value of any alternating current is equal to the value of the direct current producing the same joule heat in a resistor. To determine this value, the heating effect of an alternating current is calculated by averaging the I^2R losses over a complete cycle. For sinusoidal variation the average (active) power, P , is given by

$$P = \frac{1}{T} \int_0^T i^2 R \, dt = \frac{I_p^2 R}{2}$$

The joule heating by a direct current is $P = I^2 R$. Therefore, the effective value, I_e , of a sinusoidal current, with a peak value of I_p , is

$$I_e = \frac{I_p}{\sqrt{2}}$$

The effective value of a sine-wave current or voltage is simply its peak value divided by the square root of two. The effective value of a non-sinusoidal signal is calculated analogically. The effective value is usually termed the "rms" value (*root mean square*). For an arbitrary alternating current,

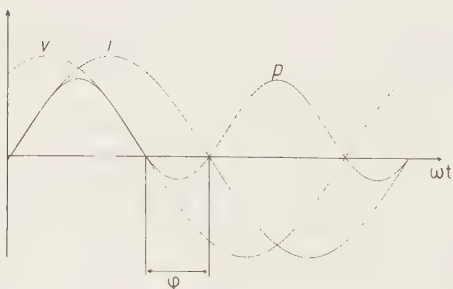
$$I_{\text{rms}} = \frac{1}{T} \int_0^T i^2 \, dt$$

In an ac circuit current and voltage may be out of phase, i.e. their rotating vectors do not coincide. Suppose, that in some element of the circuit their values are given by

$$i = I_p \sin \omega t$$

$$v = V_p \sin (\omega t + \varphi) \text{ which gives an instantaneous developed power of}$$

$$p = vi = V_p I_p \sin \omega t \sin (\omega t + \varphi) \quad (\text{Fig. 6})$$



6. Current, voltage, phase angle and power.

The instantaneous power thus varies with time and even becomes negative. The interpretation of negative power is that, during some part of the cycle, the circuit element supplies electric power to the rest of the circuit. For the remainder of the cycle, the circuit delivers power to the element under investigation.

The average power is

$$P = \frac{1}{T} \int_0^T v i \, dt \rightarrow P = VI \cos \varphi, \text{ where}$$

V and I are rms values.

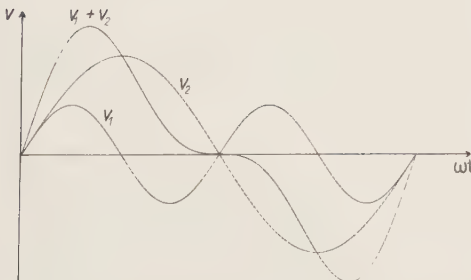
The power in ac circuits thus depends not only on current and voltage, but also on the phase difference between them. The term $\cos \varphi$, is the power factor, λ , of the circuit. In the extreme case $\varphi = 90^\circ$ ($\pi/2$ rad), $\lambda = 0$ and no active power is developed. For $\varphi = 0^\circ$, $\lambda = 1$. The oscillating component of the instantaneous power, $VI \sin \varphi$, is called reactive power.

In electrosurgery the currents are seldom purely sinusoidal. To enable circuit analysis with such complex waveforms, they have to be separated into their frequency spectrum components. Calculations can then be performed for each component separately. If we e.g. assume two sinusoidal waveforms, v_1 and v_2 , where v_1 has twice the frequency of v_2 , the resulting complex waveform is the sum of v_1 and v_2 (Fig. 7).

With established mathematical methods, expressions can be found for any periodical waveform. One way is to express the sum by dissolving it into the fundamental frequency and its harmonics (integral multiples of the fundamental). This is called a Fourier series. E.g. for a square wave of peak voltage V_p and frequency φ the Fourier series is given by

$$v = \frac{4V_p}{\pi} (\sin \omega t + 1/3 \sin 3\omega t + 1/5 \sin 5\omega t + 1/7 \sin 7\omega t + \dots)$$

The accuracy of description increases with the number of included high frequency terms.



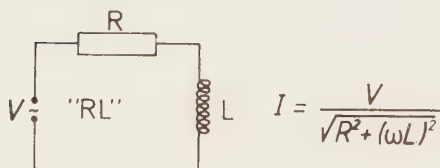
7 The complex waveform $v_1 + v_2$ and its components v_1 and v_2 .

In a circuit, where a purely resistive load is connected to a power source, the phase angle between current and voltage is zero. If the load is a pure capacitance, current leads the voltage by a phase angle of 90° , while for a pure inductance current lags voltage by 90° . The currents in RC and RL circuits are calculated as shown in Figs. 8–9. If the resistance is disregarded of, the current in a capacitance is $I = \frac{V}{1/\omega C}$ and in an inductance $I = \frac{V}{\omega L}$. The current is thus

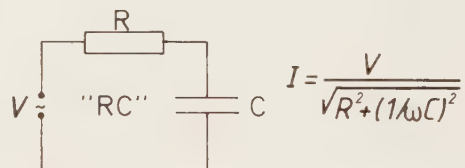
determined by $1/\omega C$ and ωL , respectively. This resembles formally the action of a resistance, where $I = \frac{V}{R}$. $1/\omega C$ is termed the capacitive reactance, X_C ,

and ωL the inductive reactance, X_L . An ideal resistance is frequency independent, while the capacitive reactance decreases and the inductive reactance increases with increasing frequency. The collective name for different combinations of reactance and resistance is impedance, and the unit of measurement is one ohm (Ω). This way of presenting reactance is inconvenient for analysis of complex circuits. If, instead, impedance is represented by a complex number, where the imaginary part is constituted by the reactance, the resulting complex impedance of a circuit is conveniently determined. The real part of the complex impedance refers to resistance. A series combination of a resistance R and an inductive reactance X_L is written in complex form as

$$Z = R + jX_L, \text{ where } j = \sqrt{-1}.$$



8. Current in an inductive circuit.



9. Current in a capacitive circuit.

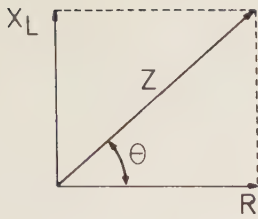
The graphical representation of this concept is shown in Fig. 10. Two complex numbers are equal only if their real and imaginary parts are equal,

thus, if $R_1 + jX_1 = R_2 + jX_2$; $R_1 = R_2$ and $X_1 = X_2$.

Two complex numbers are added by separately adding their real and imaginary parts,

$$\begin{aligned} Z_1 &= R_1 + jX_1 \\ \text{if } Z_1 + Z_2 &= (R_1 + R_2) + j(X_1 + X_2) \\ Z_2 &= R_2 + jX_2 \end{aligned}$$

The complex impedance of the inductive circuit in Fig. 8 is $R + jX_L$, while in the capacitive circuit in Fig. 9 it is $R - jX_C$. The minus sign is associated with capacitive reactance since in a capacitive circuit current lags voltage.



$$\Theta = \arctan \frac{X_L}{R}$$

10. Graphical representation of a complex number.

The electrosurgical generator: Thévenin's theorem. The maximum power transfer theorem. Significance of matching conditions

The electrosurgical generator (ESU) consists of two main components, the high frequency generator and the power amplifier. A stereo amplifier or a radio transmitter have constant loads, while the ESU is connected to a load impedance varying over a wide range. During operation, power is generated both in the external circuit and within the ESU. The generator is a complicated network of different components. A block diagram of an ESU is shown in Fig. 11. This complicated network can, if linearity is assumed, according to Thévenin's theorem be replaced by a simple equivalent circuit, as shown in Fig. 12. This allows us to determine the fraction of the power that is available in the external circuit, i.e. at the output connection, at different load impedances. The used symbols refer to those of Fig. 12. To simplify the analysis, reactances are neglected.

$$P = \frac{U^2}{R_{\text{load}}}, \text{ but}$$

$$U = E \frac{R_{\text{load}}}{R_{\text{gen}} + R_{\text{load}}} \text{ (voltage division) and thus}$$

$$P = \frac{E^2}{R_{\text{load}} \left(1 + \frac{R_{\text{gen}}}{R_{\text{load}}} \right)^2}$$

Optimum power transfer occurs for the value of R_{load} , where P assumes its maximum, that is when

$$\frac{dP}{dR_{\text{load}}} = 0.$$

Differentiation of the equation

$$P = \frac{E^2}{R_{\text{load}} \left(1 + \frac{R_{\text{gen}}}{R_{\text{load}}}\right)^2} \quad \text{with respect}$$

to R_{load} shows that

$$\frac{dP}{dR_{\text{load}}} = 0 \quad \text{if}$$

$$R_{\text{load}} = R_{\text{gen}}.$$

For the general case with complex impedances, we find, that with

$$Z_{\text{gen}} = R_{\text{gen}} + jX_{\text{gen}}$$

$$Z_{\text{load}} = R_{\text{load}} + jX_{\text{load}}$$

the conditions for maximum power transfer are

$$R_{\text{gen}} = R_{\text{load}}$$

$$X_{\text{gen}} = -X_{\text{load}}$$

which means that if one reactance is inductive, the other should be capacitive.

When the output impedance of the generator is unknown we can find it by plotting power versus load, as shown in Fig. 13. With $P_{\text{max}} = 400 \text{ W}$ for $R_{\text{load}} = 200 \Omega$ we find by insertion into the equation

$$P = E^2 \frac{R_{\text{load}}}{(R_{\text{gen}} + R_{\text{load}})^2}, \quad \text{that}$$

$$E = \sqrt{400 \frac{(200 + 200)^2}{200}} = 566 \text{ (volts rms)}, \quad \text{which}$$

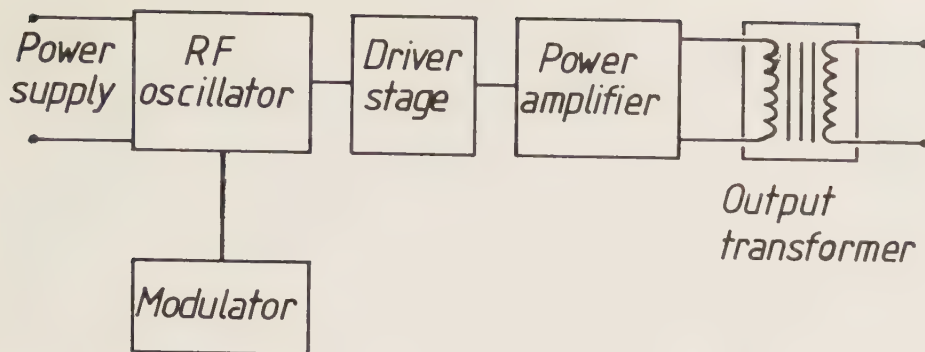
should be equal to the open circuit output voltage of the generator. Assuming a sine wave current, the peak voltage is $\sqrt{2} \cdot 566 = 800 \text{ V}$.

The power as a function of load impedance, should, if this was an ideal generator, be described by

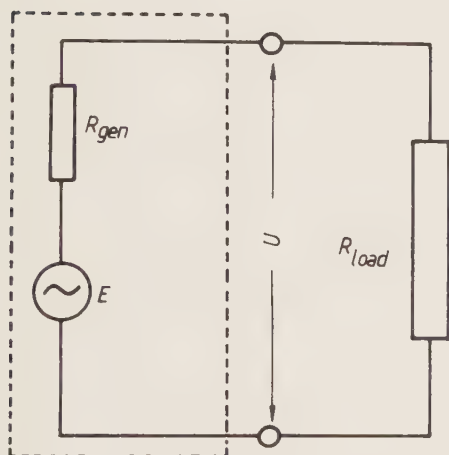
$$P = 32 \cdot 10^4 \frac{R_{\text{load}}}{(200 + R_{\text{load}})^2}$$

From this we calculate $P_{100} = 356 \text{ W}$, $P_{500} = 327 \text{ W}$, $P_{1000} = 222 \text{ W}$, $P_{2000} = 132 \text{ W}$, $P_{3000} = 94 \text{ W}$, $P_{10000} = 30 \text{ W}$.

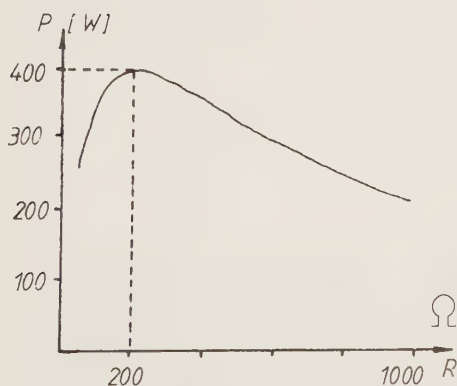
Due to non-linear properties of the electronic components, most generators may not exactly follow such equations. When the external load is equal to the output impedance, optimum matching conditions are said to exist. At loads much smaller or much larger than the output impedance, the generator is said to be working at mismatch conditions. The above computed values give an idea of the available power at different matching conditions [46, 47].



11. Block diagram of electro-surgical unit.



12. Electro-surgical unit and load circuit replaced by the equivalent circuit according to Thévenin's theorem.



13. Deduction of output impedance from power vs load plot in electro-surgical generator.

A revision of electrosurgical concepts

Surgical diathermy is a thermal aid using a primary energy source that penetrates tissue. Its power density is controlled by using electrodes of various sizes. There is only one way to produce a focusing of the energy, namely by using an electric discharge, such as a spark or an arc. A spark is a transient transfer of electric charge, while an arc is a steady state phenomenon. The concept of electrosurgical cutting as an explosive evaporation of a small number of cells into steam is attractive, since it implies that the resulting tissue damage is extremely superficial. However, the idea also has some other implications. The tissue temperature should not have to exceed 100°C if this were true. My question is then: from where does the smell of burnt flesh during cutting come? It should also be impossible to cut cell-free material if it were true. Using a soft elastomer plate, into which a saline soaked paper had been molded, I was able to produce smooth electrosurgical cutting. This material contains no cells. The developing odor was that of burning plastic. In fact, I see no reason why electrosurgical cutting should occur by mechanisms different from that of other thermal cutting. This mechanism is the rapid disintegration of tissue along a narrow line, where a high power density produces a surface temperature of several hundreds of °C. In electrosurgery this high power density cannot be produced even by the smallest electrode. No proof exists for electric discharges to be necessary for cutting, but surely no surgeon has observed cutting without arcing. The electric discharge generates power *on* the surface of the tissue due to bombardment with energetic ions and *close* to the surface because of ohmic heating by the current flowing out from the concentrated footpoint of the discharge. This is the cause of the usually superficial tissue damage occurring in cutting as well as in “fulguration”.

When hemostasis is performed by grasping a vessel with a forceps or clamp, the current waveform is completely insignificant. The action depends entirely on the average power. In fulguration a high peak voltage is necessary to draw long sparks. If the peak voltage is maintained continuously an arc would develop instead. Thus, the rms voltage is reduced by modulation of the waveform.

In applications using mechanical tissue contact, a high peak voltage at low rms voltage, may have a certain effect by producing a high current density up to some distance from the electrode. This may aid the in depth heating. If a large area electrode can be used, the same effect can be produced with a continuous current waveform. When the electrode is small, as in transurethral surgery, cutting may result also with a modulated current. The superior in depth heating with spark gap generators is related also to the load characteristics which are different from those of most solid state equipments.

Electric arcs and sparks in electrosurgery

Electric discharges have been studied for more than 100 years. However, the discharge physics still contains many unsolved problems. The electric discharges in transurethral surgery are free burning under-water discharges, driven by currents in the MHz frequency range, flowing between a small metallic electrode and biological tissue at a short, but not fixed distance.

These are all conditions which separately have been subject to little investigation. The combination has never been investigated. A spark usually means a

fast transient transport of electric charge through a gas. Its occurrence is determined by the voltage and distance between the electrodes, the electrode geometry and the gas pressure. The events after spark breakdown are determined by the characteristics of the external circuit. Electric arcs are defined by the current and the voltage drop. To maintain an arc discharge, the external power source must be capable to drive a current of some tenths of an ampère across the gap. The associated voltage drop across the gap is of the order of some tens of volts. The gas pressure affects the temperature of the arc plasma, which at atmospheric pressure may exceed 5 000 K. The frequency of the current, the composition of the gas and the electrode material are also important in determining the behaviour of the arc.

The arc discharge can be initiated in different ways, e.g. by a spark discharge. In atmospheric air the voltage required for breakdown in a gap of 2 mm, is about 10 kV. With an efficient current source, an arc is established immediately after breakdown. When the electrodes after mechanical and electric contact are separated a drawn arc results. This phenomenon is responsible for the flash often seen when an electric switch is turned off. With an adequate external circuit the arc will continue to burn.

After these general considerations on arcs and sparks we are ready to list the special features of arcs in transurethral electrosurgical cutting.

1. The arc is usually drawn. The resection loop is in mechanical tissue contact before the generator is activated. When current passes into the tissue, water evaporates and forms a vapour layer electrically separating the electrodes.

2. The gas forming the discharge plasma is water vapour of atmospheric pressure. Water vapour is a relatively unexplored gas in plasma physics. Literature and investigations (Lundquist, S.: personal communication 1981) indicate that water vapour can be highly efficient for extinguishing arc discharges!

3. The arc is free burning and surrounded by water. Most arcs studied in the laboratory have been stabilized, e.g. by confining them to a tube with cooled walls. The free burning arc is surrounded by a thermodynamic boundary layer functioning as a boundary condition of the discharge. This layer is, however, stationary only under certain conditions. One condition is that the electric power must be slowly varying. Furthermore, only a small part of the heat exchange should be due to induced gas streams in the boundary layer. The short gap distance in TUR favors the arc stability by reduction of heat convection. However, the thermal boundary conditions are extremely unfavorable because of the gas/water interface, which has a higher heat transfer coefficient than gas/gas. Increased input power disturbs the flow pattern around the current channel, which in turn causes movements of the arc channel. The low current arcs encountered in TUR are for the same reasons more susceptible to interruption and instability than high current arcs.

4. The electrode composition is complex. The voltage drop in short arcs occurs almost exclusively close to the electrodes. From these regions emission of electrons/ions must take place to enable the charge transport. Investigations which simulate the TUR conditions concern discharges between a carbon electrode and a metallic electrode. The temperature at the metallic electrodes in these cases is high enough to produce jets of vaporized electrode material, which are likely to interrupt the current leading to a creation of new electrode spots. This causes the footpoint of the arc to move rapidly over the electrode surface. These phenomena are at least partly frequency dependent.

5. The frequency of the current is in the MHz range. Our choice of frequency is limited by hazardous physiological effects at lower frequencies and technical transmission problems at higher frequencies. I will in spite of this mention the frequency dependence of arcs. Below about 100 Hz, the ionization of the gap disappears during each half cycle at the passage of current zero, with consequent extinguishing of the arc. At some kHz the gap remains ionized during the entire cycle. The arc will in much behave like an ohmic conductor upwards to some MHz. Above that, changes of ionization and diffusion due to altered electron and ion mobilities affect the arc characteristic.

6. The temperature of the arc. The central temperature of stable arcs at atmospheric pressure usually approaches 5 000 K.

7. The electrode gap is very small. Few investigations concerning discharges in very small gaps have been performed.

8. One electrode is moving during cutting. This fact reemphasizes the necessity to create electrically optimized conditions for a stable arc. As it appears, the entire loop is smoothly cutting through the tissue. This is unlikely to occur with a single arc. It would then have to move so fast along the electrode, that it is not probable that the plasma would be maintained, or the electrode would have to be moved very slowly which would increase the thermal load on the tissue. It thus seems probable that several arcs should be available at the same time. The scientific evidence may come from an investigation which is underway. High speed films and synchronized recordings of transient current and voltage changes must be obtained. The movement of the electrode has another important implication. If it is not surrounded by a perfect film of water vapour, the change in load impedance presented to the power source is likely to produce rapid changes in power, which may disturb the stability of the arc. It depends entirely on the power source (the ESU) if this is the case. The proposed idea to limit the discharge to one arc, by regulating the current from the power source, is therefore questionable.

The conclusions of this discussion are

- a) The electric function of the different components of the entire electrosurgical circuit should be analyzed
- b) The load characteristics of electrosurgical generators should be investigated and correlated to their surgical performance
- c) Measurements and recordings of currents in TUR should be performed, as well as investigations of the loads during cutting
- d) Cutting requires a stable and relatively high current output to produce arcs
- e) Fulguration requires low average current but high peak voltage to produce sparks
- f) Desiccation and coagulation should be performed without electric discharges.

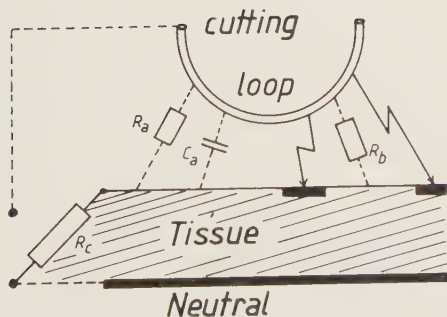
Literature [48–54]

Electric equivalents to electrosurgical cutting in TUR

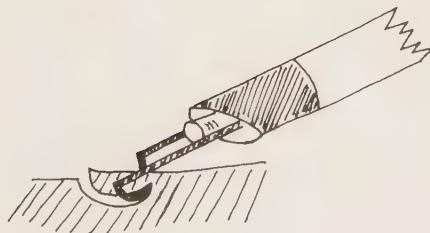
The circuit in electrosurgery consists of

1. The electrosurgical unit
2. The “surgically inactive” part of the patient, which constitutes a constant impedance
3. The “surgically active” region.

A schematic of the circuit is shown in Fig. 14. R_a is the transitional loop to tissue resistance, which is in parallel with the capacitance C_a , representing the capacitive coupling of the electrode to the tissue. R_b is the resistance of the electric arc(s). R_c is the resistance of the patient current path to the neutral. The transitional resistance of the neutral is also included in this term. Several other reactances which exist in the circuit are difficult to determine. They have been omitted, though they may to some extent affect the active power by producing a phase difference between current and voltage. If we first regard the moment when the generator is activated, we find that the load consists of R_c , R_a and C_a . The contact area between the electrode and tissue is rather small, and the resulting R_a will be large. The impedance presented by C_a is also large. Current passing into the tissue causes heating and evaporation of water along the entire part of the loop in tissue contact. A film of steam insulates the loop from the tissue. The result is one or more drawn arcs. Under these conditions, C_a and R_c are assumed to remain unchanged, while R_a approaches infinity. The upper limit of the voltage between the electrode and tissue is set by the smallest of the parallel impedances. The arc can be expected to have a very low resistance when established. The resistance between the tissue foot-point of the arc and the underlying tissue may change during the cut, because tissue resistivity is changed by heating. The situation is even more complicated by the fact that in loop resection, a chip of tissue is gradually raised by the loop. The bridge to the remaining tissue thus becomes small towards the end of the cut, which could be expected to increase the total resistance, partly due to a decrease of cross-sectional area of the tissue allowing for current return, partly due to a decrease of the contact surface area of the loop to tissue (Fig. 15). The



14. Schematic electric circuit corresponding to electrosurgical cutting.



15. Loop resection.

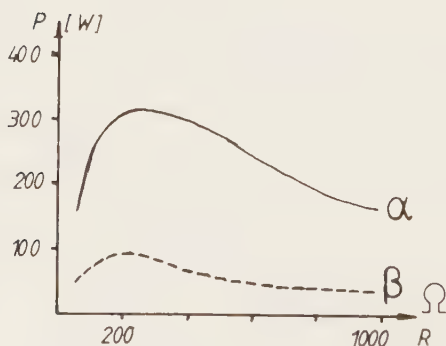
result of these complicated events is hardly expected to be the same in different cuts. The technique of the surgeon will also have an influence. However, the relevant question is, if the changes of load impedance in the vicinity of the electrode are significant compared to the constant part of the load, R_c , to which the electrode-tissue resistance is connected in series. The answer is crucial, because it determines the change in the load impedance, seen by the electrosurgical generator during the cut. If matching conditions are significantly changed, the current delivered by the generator may vary considerably. This affects the stability of the arc, and may even extinguish it. The total impedance therefore has a critical effect on performance.

A comparison of load characteristics with surgical performance in some electrosurgical units

Load characteristics have previously in general not been specified by manufacturers. There is hope that this will become more usual, because of the recommendations of IEC 62D (CO) 5 and IEC 62D (CO) 12 (International Electrotechnical Commission), and ECRI in Health Devices. IEC recommends loads of 50–2 000 Ω , while the ECRI recommendation is for loads up to 1 000 Ω . As long as the measurements are not performed in the same way, there is a risk for misleading or erroneous results.

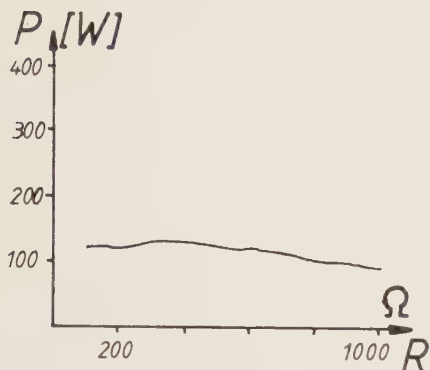
Usually such measurements are performed as current measurements with a radio frequency ammeter with power calculated from the value of the known load resistor. The resistors that are commonly used are non-ideal at diathermy frequencies. This means that a reactance may have been added to the circuit. The reduction of active power resulting from this is not detected, since only current is measured. The error is small, compared to what may result from erroneous loading of the generator. If the smallest load impedances are used first, the unit may be overheated and give less power than normal at the higher loads. An example of this error (own graph) is shown in Fig. 16.

Since it was impossible to have all the generators available at the same time for measurements and since specified characteristics existed only for a few of them, the presented data are partly taken from available literature [39, 55], partly from own measurements.

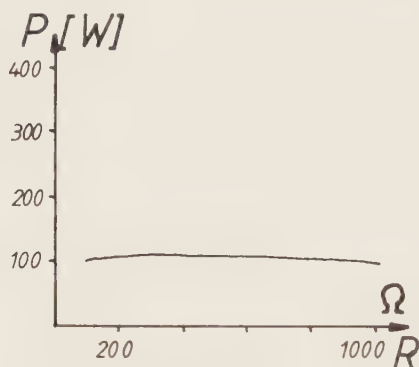


16. Erroneous power vs load plot. In α loading was started at low values. The heat generation in the ESU caused a decrease in power output. In β measurements started at the higher loads.

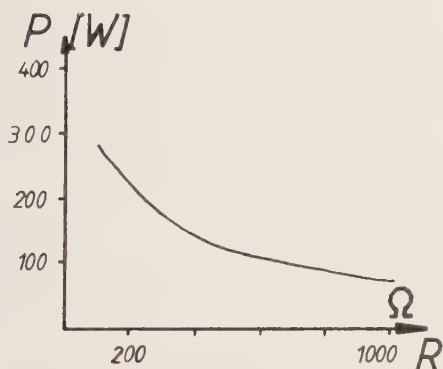
The generators chosen for the comparison have all been used clinically at our department, and their surgical performance is well known. The load characteristics and comments on the corresponding surgical performance are shown in Figs. 17–24. Figs. 25–33 show the characteristics of three generators with assumed properties, computed and plotted from the “generator equation”.



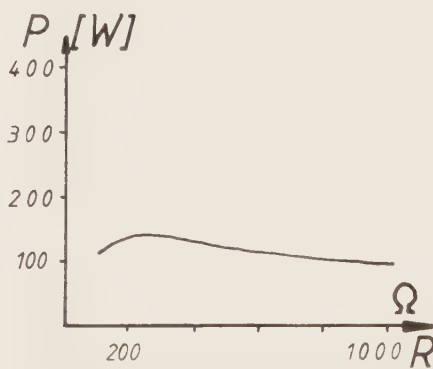
17. Spark gap generator. COAG mode. Excellent coagulator.



18. Power regulated solid state unit. COAG. Excellent coagulator.



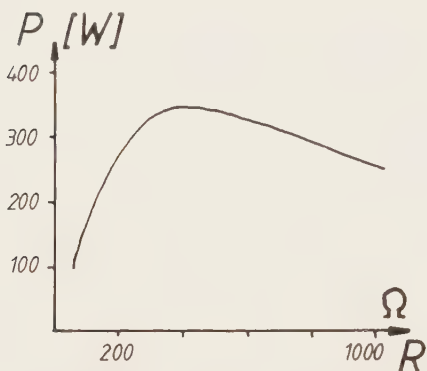
19. Conventional solid state. COAG. Poor coagulator, marked cutting tendency.



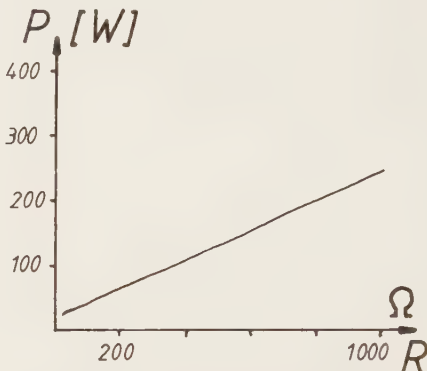
20. Conventional solid state. COAG. Hemostasis performance somewhat less than fair.

The conclusion is that the two excellent coagulators, one power regulated solid state equipment (Fig. 18) and a spark gap machine (Fig. 17), both prove to be constant power generators. The poor coagulator (Fig. 19) has a characteristic more resembling the CUT mode graphs than that of the fair coagulator (Fig. 20). The two excellent cutters, one solid state (Fig. 21) and one tube generator (Fig. 22) are not obviously different in their power versus load characteristics, but if current is plotted versus load, both show small changes in current at loads up to 1 000 Ω . The tube generator actually works much like a constant current generator, which also agrees with its high output impedance, probably being about 100 k Ω . The solid state generator (Fig. 21) delivers more than 1 000 W at maximum power setting and optimum match! At the setting used for plotting the graph it gave somewhat higher current readings than the poor cutters (Figs. 23–24) at maximum power setting. The truth is probably revealed if the change of current per unit load impedance change at higher loads is compared

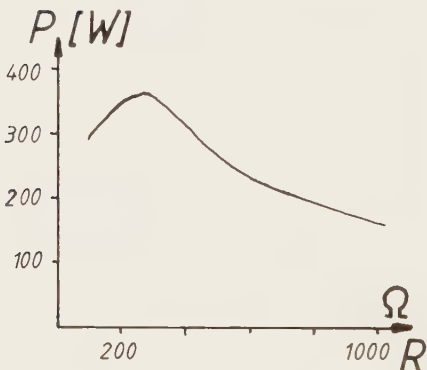
for the two generators. One of the conditions for a stable arc, was that current changes across the gap should be small and occur slowly. This makes it reasonable to expect better cutting performance from a generator of the constant current type [56, 57].



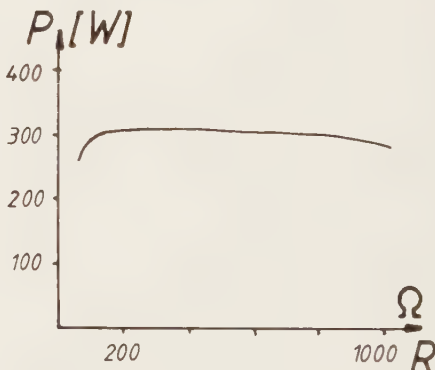
21. CUT 4/10. Solid state. Excellent cutter. Maximum power exceeds 1 000 W.



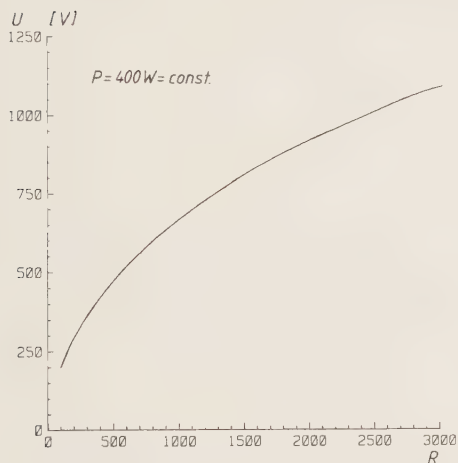
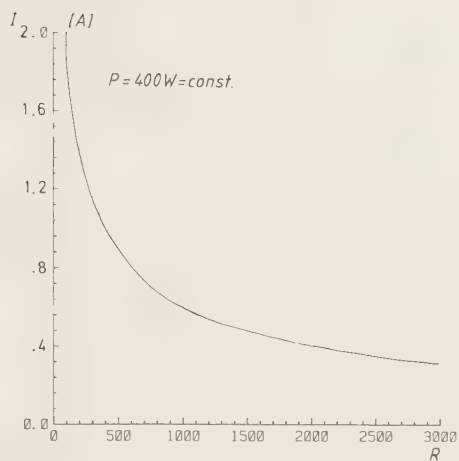
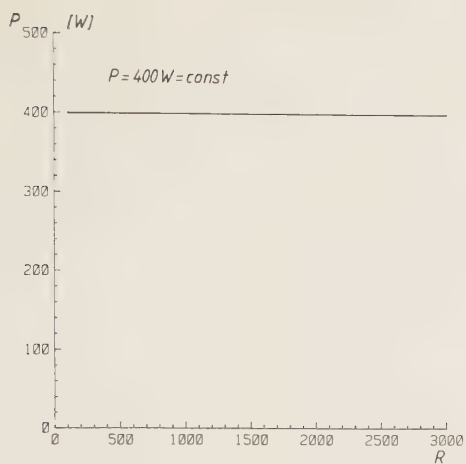
22. CUT max. Electronic tube generator. Very even but somewhat slow cutting.



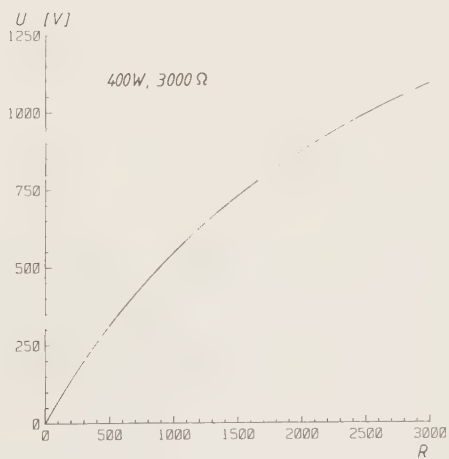
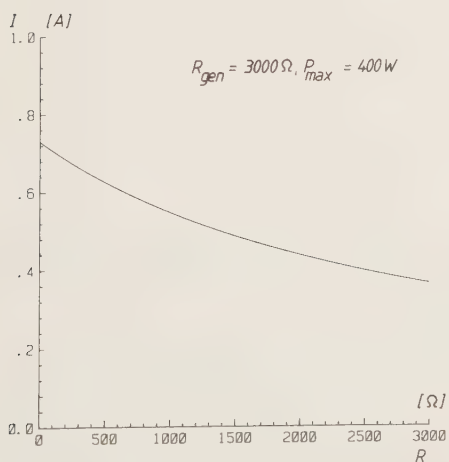
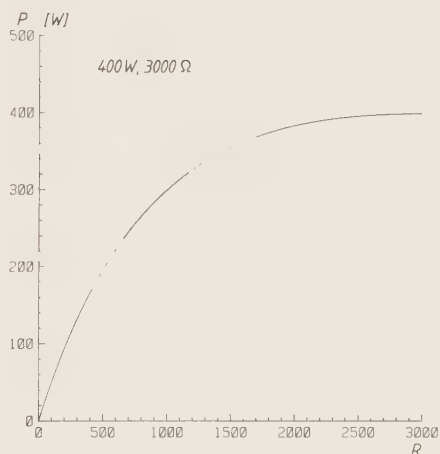
23. CUT max. Solid state. Extremely poor and uneven cutting performance.



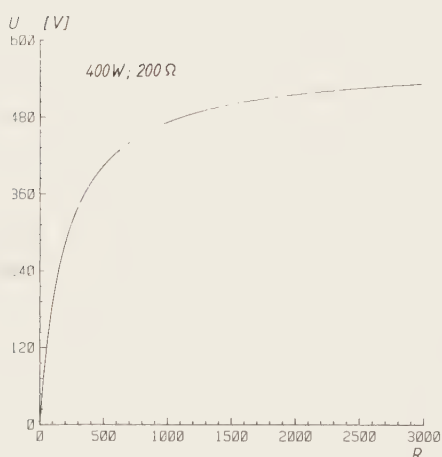
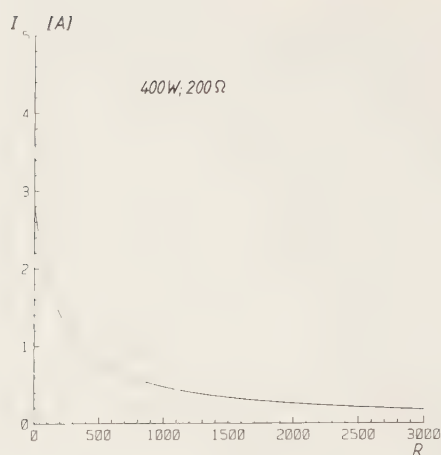
24. Power regulated solid state unit. CUT max. Extremely poor cutting unless power setting is close to maximum.



25–27. Power, current and voltage versus load. Constant power generator. Computed.



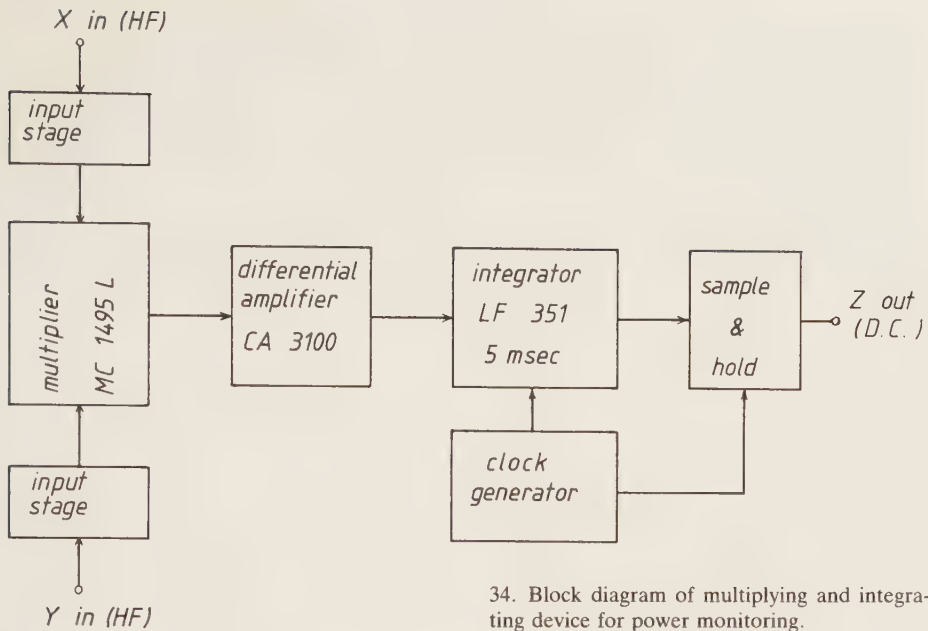
28–30. Power, current and voltage versus load. High output impedance. Computed.



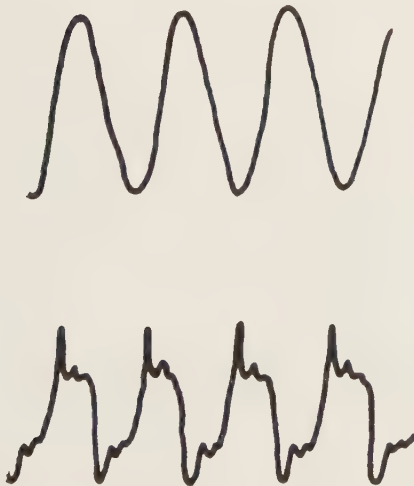
31–33. Power, current and voltage versus load. Ordinary output impedance for solid state generator. Computed.

The construction of a device for measurement of current. Recordings during experimental electrosurgery

Our original intention was to record the power during cutting. This can only be performed by simultaneous recording of current and voltage. Several possibilities exist for recording. We decided to use a multiplying circuit to obtain instantaneous current, voltage and power readings. By integrating the values over 5 ms and converting the signal into a dc voltage proportional to the input, a simple chart recorder could be used. A block diagram of the device is shown in Fig. 34. The current readings were obtained from a current transformer

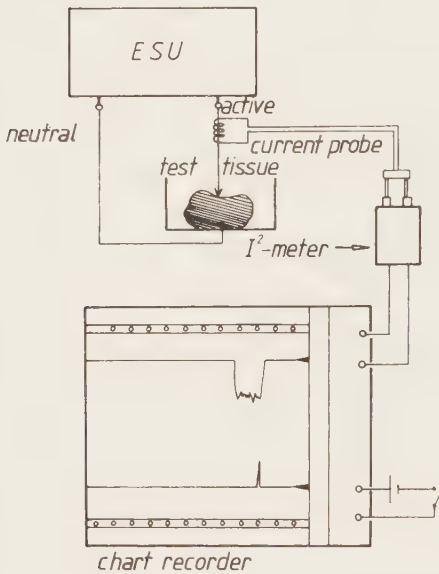


probe (Tektronix P6022). For voltage readings, a high frequency attenuating voltage probe (Tektronix P6015) was used. However, it proved impossible to adjust the compensation for this probe, due to the severe harmonic distortion of the sinusoidal waveform. The difference between the distorted and undistorted waveforms is shown in Fig. 35. This became obvious at mismatch, and



35. Above undistorted sine wave. Beneath the appearance at mismatch.

made measurements impossible. Thus we had to resort to connecting the current probe to both inputs of the multiplying unit. The resulting dc signal thus became proportional to I^2_{rms} . A dual channel chart recorder (Servogor) was used for recordings. Its second channel was arranged as an event marker. In the laboratory tests of this measuring procedure and equipment we followed the advice given by manufacturers for getting acquainted with new electrosurgical equipment. The suggestion was to place a piece of raw meat on the neutral and test cutting. Our experimental arrangement for cutting and recording is shown in Fig. 36. Three different electrosurgical units were used. One



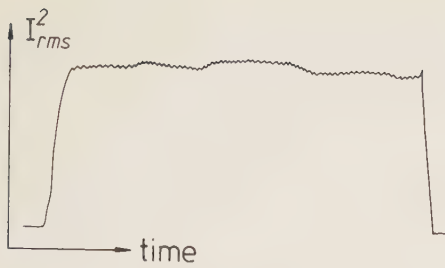
36. Arrangement for current recordings during experimental electrosurgery.

was an electronic tube generator (Wappler C-263, made ~ 1950), the other two solid state generators. One had a poor clinical cutting performance (Philips TS 403), while the other was excellent clinically (EMS 2000). Only one unit produced fair cutting in this situation, namely the tube generator. Some of the recorded current graphs are shown in Figs. 37–42.

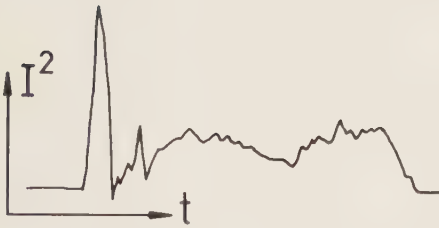
Recordings were taken during cutting with different electrode types, in air and under water.

In the fair cuts that were produced, the current was nearly constant. Sticking of the electrode to the tissue was associated with a sharp rise of the current. A few coagulation recordings were also performed, only two, however, under fully controlled conditions.

Literature on electronic measurements [58, 59]



37. Tube generator. Smooth cut.



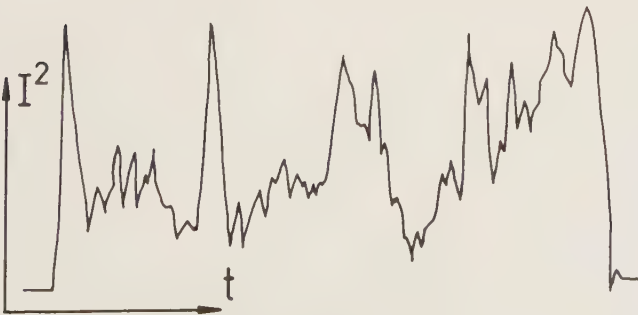
38. Solid state generator. Fair cut, electrode in tissue contact before the generator is activated.



39. Solid state generator. Fair cut, generator activated before the electrode contacts tissue.



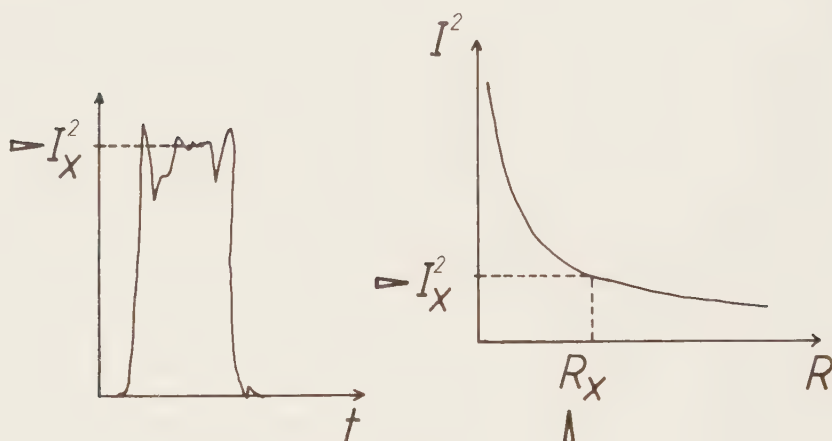
40. Solid state generator. No cutting occurred. Heavy sparking and charring.



41. Poor and uneven cut with solid state generator. The electrode is stuck to tissue at the current peaks.



42. Solid state generator. Smooth cut.



43. Load determination from current recording and load-current characteristics of generator.

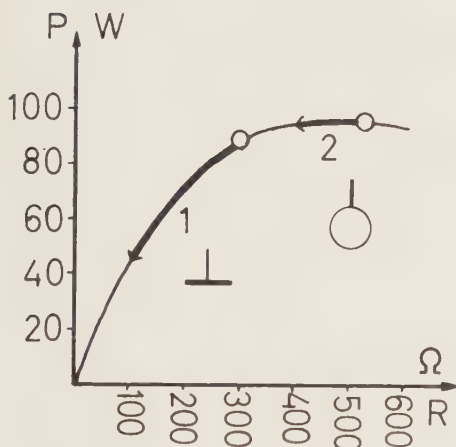
Determination of load impedances corresponding to the recorded currents

For the used generators and power settings, I_{rms}^2 versus load was plotted. These measurements were performed with large carbon resistors, with a reactance less than $1 \mu\text{H}$, and negligible temperature dependence. For each I_{rms}^2 -value recorded during cutting it was thus possible to find the corresponding load (Fig. 43). The initial reading is likely to be about equal to the sum of R_c and R_a (Fig. 14).

With a needle electrode in air the initial load was in the order of $1\ 200 \Omega$. Wire loops (diam. 10 mm) gave initial loads of $600\text{--}800 \Omega$. Under water the initial loads were about 200Ω lower. Smooth cuts in air with the loop produced loads of $1\ 000\text{--}2\ 000 \Omega$, while under water the range was $1\ 500\text{--}2\ 500 \Omega$. In the uneven cuts, with sticking of the electrode, the impedance varied between 600 and $2\ 000 \Omega$ within less than 0.1 sec. The corresponding currents were $0.5\text{--}1\text{A}$.

The coagulation experiments were performed under water with a large button electrode (area = 1 cm²) (1) and a 10 mm diam. wire loop (2) and the generator set to a low power CUT mode. The results in terms of impedance and power are shown in Fig. 44. No charring or evaporation occurred. It is generally assumed that tissue resistivity and hence impedance increase at coagulation. Our results indicate the opposite. Other investigators have found the same to be true, until the tissue is entirely desiccated, when impedance rises rapidly. When disintegration occurs, the impedance once again decreases (personal communication; Vällfors & Bergdahl, 1981). The resulting increase in power occurring with an ordinary solid state generator is a possible explanation of the cutting that may result during coagulation.

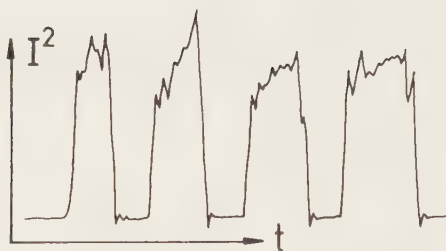
By the combination of a loop and a ball electrode we also demonstrated the feasibility of bipolar cutting.



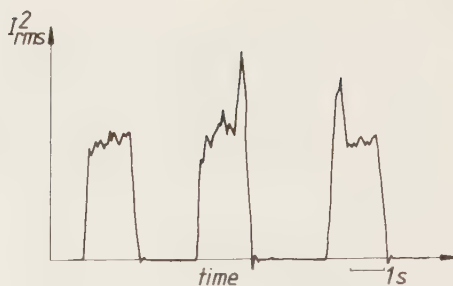
44. Impedance variation during coagulation. Plotted into power vs load graph.

Preliminary clinical investigations on currents and load impedance during TUR

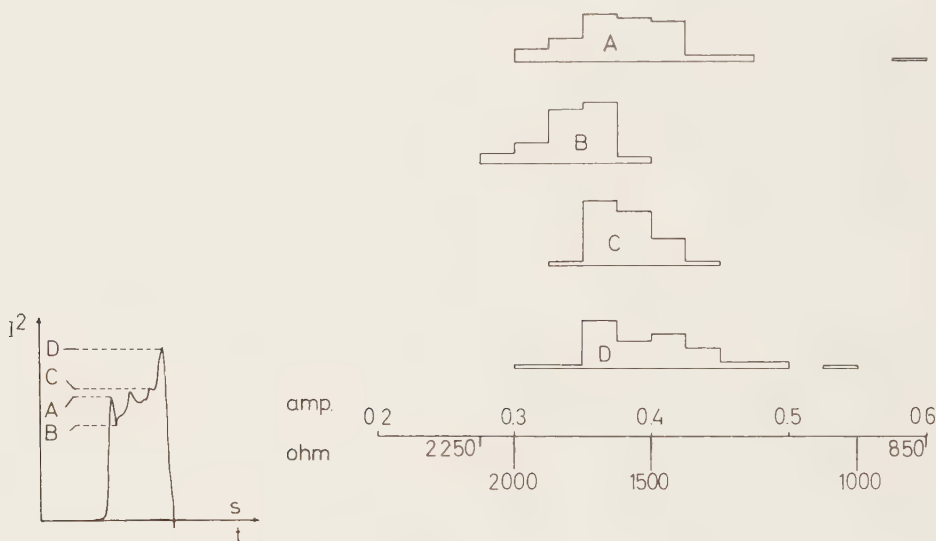
The measurements were performed in the way described above. The current transformer probe was clipped onto the active lead. No electric interference with the electrosurgical system was thus needed. Disturbance of the measurements might have been expected in the electrically complicated operating room environment, but did not occur. The used generator was an EMS 2000 BP, set to a pure cut mode. Recordings during coagulation were discarded. The patient suffered a WHO stage I prostatic carcinoma and benign prostatic hyperplasia with obstruction. The procedure was performed by an experienced resectionist. Intermittent irrigation with 22 mg/ml glycine solution was used. The resectoscope was a Storz 27F. Cutting was estimated excellent throughout the procedure. We determined the duty cycle to average 65%, with a maximum of 75% for single 5 min periods. A total of 45 cuts were recorded. A typical series of cuts is shown in Fig. 45. Some extreme forms are shown in Fig. 46. The distribution of initial current, low and high values during cutting and current at termination are presented in Fig. 47. The main variation range was 0.3–0.45 A, with a few extremes above 0.5 A. The initial load median was found to be 1 600 Ω.



45. Current during a series of smooth cuts in TUR.



46. Some extreme forms of current graphs during TUR.



47. Distribution of current and load impedance in 45 cuts during TUR.

During the cut the load varied between 1 200–2 200 Ω , with a median value at 1 650 Ω . The median load value at termination of the cut did not differ significantly from that during the cut.

The over all extremes of variation were 850 and 2 500 Ω . The low extremes were recorded during termination of the cut, probably due to tissue chips sticking to the loop and being withdrawn into the resectoscope sheath.

The output power was in the 180–250 W range.

Concluding remarks

The total load impedance varies between 850 and 2 500 Ω , which means that a solid state electrosurgical unit is working at considerable mismatch. The transient changes in the load put high demands on the generator. Since most generators, when working at a power setting close to maximum, produce large changes in delivered power in response to a minor change of load, it is not surprising to find that a generator capable of power output exceeding 1 000 W at 400 Ω is a better cutting device than one capable of 400 W at an optimum match of 200 Ω . It should be noted that this is not because of the higher available power, since about the same power is produced in cutting with the less powerful unit. It is probable that the slope of the current versus load graph in the range of 500–3 000 Ω is the relevant parameter, i.e. the closer to horizontal the graph is, the more even is the cutting. In fact this means getting closer to the characteristics of a constant current generator. The discussed conditions for a stable arc discharge also reinforce this assumption. The poor performance of the generators in the laboratory tests is also explained by the small initial load which causes the variations during cutting to produce a proportionally larger change of load. The working load range was also located at the steepest downward slope of the power vs load graph, meaning that a relatively small change in load resulted in a major change in available power (cf Fig. 44). Specifications of power/current-load characteristics, in the range of loads up to at least 3 000 Ω should therefore be requested from manufacturers. This is the only objective information that indicates the surgical performance of electrosurgical units.

As a summary, these investigations point at several possibilities of improving electrosurgical equipment.

Additional literature [60–74]

Development of an invasive microwave heating method for transurethral in depth coagulation of tumors

The aim of this work has been to investigate the feasibility of using an invasive microwave antenna as a surgical instrument. The therapeutic goal is the in depth destruction of infiltrating bladder carcinoma. Since this particular application of microwaves has not been reported earlier, the investigations have been guided by theoretical considerations combined with experiments.

Basic electromagnetics. Coupling of em energy to tissue

The laws of electricity are summarized in Maxwell's equations, which imply that a time varying electric field is accompanied by an magnetic field and that a time varying magnetic field is accompanied by an electric field. At high frequencies electric and magnetic fields are strongly coupled in the form of electromagnetic waves. A plane wave consists of an electric field (E) and a magnetic field (H), mutually perpendicular and confined to planes perpendicular to the direction of propagation. The travelling wave represents a transport of energy, described by the Poynting vector $S = E \times H$ (Wm^{-2}). The propagation velocity is determined by the relation

$v = \lambda \nu$ (ms^{-1}) where

ν = the frequency (cycles per second, Hz)

λ = the wavelength (m).

The propagation velocity and the wavelength are determined by the electric and magnetic properties of the medium.

Microwaves have frequencies in the GHz range and wavelengths in vacuum from 1 to 100 mm.

In vacuum the propagation velocity is $G_0 = 2.998 \cdot 10^8$ (ms^{-1}). The electric properties of a medium are described by its permittivity, which can be expressed as a complex number $\epsilon = \epsilon_0 (\epsilon' - j\epsilon'')$, where ϵ_0 = the permittivity of vacuum (Fm^{-1}).

The magnetic properties are described by the permeability μ , which can also be expressed by the relative permeability, $\mu_r = \mu/\mu_0$, where μ_0 is the permeability of vacuum. For biological tissue, $\mu \approx \mu_0$. (Hm^{-1}). Tables describing the dielectric properties of tissue are available in the literature. Often the conductivity corresponding to a given frequency, σ (Sm^{-1}), is presented together with ϵ' . σ is related to ϵ'' and the frequency f by $\sigma = \omega \epsilon_0 \epsilon''$, where $\omega = 2\pi f$ is the angular frequency.

The interaction of electromagnetic waves with biological tissue can be considered as an effect of the electric field. Energy is exchanged with free charges (ions) and with molecules having an asymmetrical charge distribution (electric dipoles). Charge redistribution under the influence of the field causes the induction of dipoles. These tend to be aligned with the existing field. Changes in the orientation of dipoles lead to their rotation in the alternating field. The rotation of dipoles contributes to the displacement current through the dielec-

tric, whereas the oscillation of free charges gives rise to conduction currents. The mechanical work exerted by these processes is the basic mechanism behind the thermal effect of electromagnetic waves. The loss tangent, $\tan \delta = \frac{\epsilon''}{\epsilon'} = \frac{\sigma}{\omega \epsilon_0 \epsilon'}$ is used to express the frequency dependant energy loss in a medium.

The amplitude of a plane wave penetrating into biological tissue decays exponentially due to energy absorption in the tissue as expressed by $A(x,t) = A_0 e^{-\alpha x} \sin(\omega t - \beta x)$, where

A_0 = the maximum amplitude at $x = 0$ (the tissue surface)

A = the amplitude in the plane x at the time t

α = the attenuation per unit length

β = the phase shift per unit length.

The absorbed power per unit volume at any point is

$$q = \frac{1}{2} \sigma |E|^2 \quad (\text{Wm}^{-3}), \text{ where}$$

$|E|$ = the absolute (peak) value of the electric field (Vm^{-1}).

At a given frequency the characteristic depth of penetration is described as $d = 1/\alpha$, at which the absorbed power density has decreased by a factor $e^{-2} = 0.13$. Since a major part of absorption at microwave frequencies depends on the rotation of water molecules, the concentration of water in tissue is important in determining the dielectric losses. The wavelength λ for an electromagnetic wave with the frequency ν in a medium with

$\epsilon = \epsilon_0(\epsilon' - j\epsilon'')$ and $\mu \approx \mu_0$ is given by

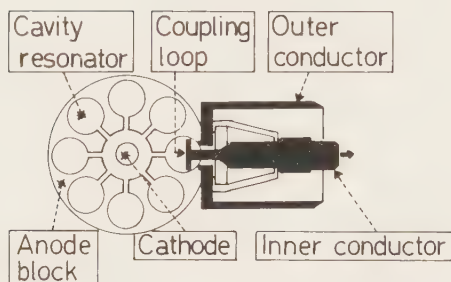
$$\lambda = C_0(\epsilon')^{-1/2} \cdot \nu^{-1}.$$

In the bladder wall the internal tissue layer is comparable to muscle tissue, while the external layer consists mainly of fat. Some values for these tissues, are presented in Table A.

For generation of electromagnetic waves a high frequency generator is required. If the generator is connected to a closed oscillating circuit very little radiation is generated since the contained energy is alternately stored as electrostatic energy and magnetic energy. An oscillating and radiating circuit is a simple antenna, where a wire represents a self-inductance and the capacitance is distributed between different parts of the wire and its surroundings. The radiation can be ascribed to the open geometry of the electric field. The design of the generator changes with power and frequency. For low power requirements in the microwave range (e.g. hyperthermia applications) semiconductor circuits can be used. At the lower frequencies electronic valve devices can be used. For high power, the magnetron is at present the most economic microwave source. The first low power magnetrons were developed just before World War II. During the war, the development of radar techniques for military purposes made magnetrons unavailable to medicine. A magnetron consists of a cylindrical vacuum tube and a permanent magnet (Fig. 48). In the center of the tube is a heated cathode. The anode consists of a cylindrical block, usually made of copper with the inner surface broken by a series of resonant cavities. The magnetic field is perpendicular to the anode. A voltage is applied between the electrodes. Electrons emitted from the cathode are

accelerated radially towards the anode, but are deflected by the magnetic field. Their resulting trajectory is a cycloidal path along the interelectrode space. The motion of the charges, by resonance, builds up oscillations of high amplitude and high frequency in the resonator cavities. The electromagnetic energy is extracted from one of the cavities by an inductive loop and conducted to the radiating device by means of a rectangular or coaxial waveguide. A modern magnetron is a highly reliable generator of microwave energy.

Literature [75–78].



48. The magnetron.

The penetration depth is given by

$$d = 1/\alpha = \frac{C_0}{\omega} \left[\frac{\mu_r \epsilon'}{2} \left(\sqrt{1 + \left(\frac{\epsilon''}{\epsilon'} \right)^2} - 1 \right) \right]^{-1/2}$$

Values of d for some frequencies are listed in Table B. From these figures it is obvious that a controlled deep heating is possible only at radio frequencies (below 300 MHz). The simplest way of coupling the energy into the tissue is to use a radiating waveguide aperture. One of the associated problems is that an efficient radiator must have a dimension of about one half wavelength in the medium it is designed for. As an example, the wavelength at 10 MHz in muscle tissue is 1.2 m. The radiator, thus, should have a transverse dimension of about 0.6 m. If it is made smaller, it does not radiate. The cut-off frequency can be lowered by loading the waveguide with a proper low-loss dielectric. Such waveguide applicators have been designed for 27 MHz. Their size limits the number of possible applications. If only a superficial heating is required, the same approach can be used at microwave frequencies.

At radiofrequencies, the energy can be coupled by two electrodes, one on each side of the part of the patient, which is to be heated. A high E-field is produced perpendicular to the electrodes. The problem with this application is, that a strong heating of low conductivity subcutaneous fat occurs. The thermal load adjacent to each plate can be lowered by using more than two plates in a "cross-fire" arrangement [79]. The size of the plates limits the clinical usefulness. As an alternative inductive applicators can be used to couple an H-field to the tissue. Heating then occurs by the induced E-field. At microwave frequencies this approach is impractical, because the heating will be very superficial. One commercially available system, which uses a large coil surrounding the patient, operates at 13.56 MHz (80). Preliminary clinical experience is promising.

None of the discussed modes is useful for endoscopy, since this application requires a small size radiator as well as transmission line. Furthermore, antennas in a biological tissue, will not act as conventional radiating systems. They must be regarded as electrically small. As a consequence of this fact and because of the wave attenuation in the lossy dielectric a useful far field, i.e. proper radiation, cannot be achieved. This is fortunately not a disadvantage in surgical applications.

Additional literature [81–82]

Previous reports on invasive microwave applicators and surgical use of microwaves

Only a few groups have been working with invasive application of microwaves, all aiming at conventional hyperthermia.

The "microwave syringe" concept was reported by Taylor in 1977. Development of the necessary technique resulted in its clinical use in 1980. Surgically inaccessible glioblastomas have been treated with a surgically implanted antenna [83]. The device which operates at 2 450 MHz is described as an open-circuited insulated short section of a transmission line, made from a coaxial cable, with the inner conductor extended "a proper length". The radiation is described as that from a distributed current source. The major heating occurs by deposition of energy in the near field zone. The heating pattern has been investigated by the split phantom technique after a 30 s heating at 50 W, using a thermographic documentation. Experimental in vivo hyperthermia in feline brain with thermal mapping at steady state has also been performed. About simultaneously, Bigu-del-Blanco and Romero-Sierra [84] described experimental work with a "monopole radiator", operating in the X-band (~ 10 GHz). No investigation of heating patterns was performed. The design of the equipment seems to limit the clinical usefulness. Douple, Strohbehn a.o. have reported on an invasive antenna intended for hyperthermia treatment by insertion into tumors. The antenna was originally described as a half-wave dipole, later as a quarter-wave monopole. Mathematical solutions for the heating pattern have been obtained, as well as experimental determination of steady state, temperature profiles [85]. The original antenna consisted of a semirigid coaxial cable, with the bare inner conductor extended about 11 mm. Later, an investigation of different antenna designs with regard to insulation etc. was reported [86]. Since the total range of allowable heating was considered to be 37–45°C, and the therapeutic range 42–45°C, a parameter " $(r)_{S_{\Delta T/8}}$ ", was introduced to

describe the steady state performance of the antenna. Studies were performed at 1–4 GHz. The effect of hyperthermia induced with this antenna in experimental animal tumors has been investigated. To our knowledge clinical experience has not yet been gained. The approach to the associated problems is purely scientific. Mendecki Friedenthal, Sterzer a.o. have published numerous reports on an invasive applicator operating at 915 or 2 450 MHz, mainly intended for transrectal hyperthermia induction in prostatic carcinoma. The antenna is described as a half-wave dipole. It is made from a coaxial cable, with the inner conductor extended an unspecified length. For the transrectal application, the radiator has been encased in an asymmetrical teflon bulb, which produces a “directed radiation field”. The temperature distribution in the canine prostate has been examined. In 1980 the treatment of two patients was reported [87]. The treatment was performed in conjunction with ionizing radiation and with RF-induced hyperthermia, which makes the evaluation impossible. Both patients were stage T4,M1. The clinical indications for the treatment and the results have not yet been reported. Furthermore, the received 43–45 Gy of radiotherapy alone can be expected to have caused the observed local tumor regression.

In 1980 an invasive device was reported by Szwarnowski and fellow investigators [88]. The special feature was a retractable inner conductor. Their method of recording the heating pattern must be considered rather inexact. A radiator of entirely different design has been described by Petrowicz and co-workers (89). Its physical dimensions exceed those of the other described radiators (20 mm diam.), and it is designed for transrectal hyperthermia treatment of prostatic carcinoma. The radiator is described as a cylindrical slot with the outer cover water-cooled to 10°C. In a series of experiments with heating of the canine prostate it was obvious that heating must have been caused by the far field (radiation). The operating frequency was 433 MHz. The temperature measured in the urethra was between 44 and 48°C, which indicates the presence of a higher temperature in the posterior prostate. This is also indicated by histological examinations revealing necrosis. Clinical use was considered, but had not commenced in June 1981.

Most of these applicators require an external matching network. The small volumes (≤ 20 mm diam.) accessible to this form of hyperthermia application indeed makes the indication questionable, except for brain tumors. Most other tumors of this small size can be removed by surgery. If invasive microwave technique is to be used, the selected temperature must cause a secured therapeutic effect in order to be justified. The heating of prostatic carcinoma by this modality is also questionable. In the case of a locally extensive tumor growth, causing local pain, the preferred modality should be regional, rather than local hyperthermia. This should be used in combination with a dose of ionizing radiation, lower than that otherwise used, if any therapeutic gain is to be expected.

Additional literature [90–94]

The first report on microwaves as a possible aid in surgery was published in 1961 by Linke and co-workers [95]. In their investigations kidneys and portions of the liver in rabbits were exposed to a 2 450 MHz field radiated from a corner reflector. The exposure times were long. No temperature measurements were made and large volumes of necrotic tissue were produced. The vital structures

adjacent to the irradiated organs were shielded by aluminum foil. After a number of comparative studies of heating with surgical diathermy, actual cautery and hot water [96], Linke and Magin in 1980 investigated the possibility of destroying the canine prostate by intense microwave radiation [97]. The organ was exposed surgically. With adjacent bowel carefully shielded, heating was produced by a 2 450 MHz direct contact waveguide applicator. The power was controlled to maintain a temperature of 60°C at the point most distant to the source. This temperature was reached in about 10 minutes, and was maintained for 15 minutes. The temperature of the organ surface adjacent to the applicator exceeded 90°C. Histology within the first week showed massive thermal necrosis. The animals tolerated these large necrotic masses surprisingly well. The urinary outflow had been secured by urinary diversion. Histology at 6 months showed fibrosis with no residual prostatic tissue. It should be pointed out, that this approach is impossible in man, because of the anatomical difference.

In 1978 Osaka reported a 2 450 MHz high power irradiation in a free field (= microwave oven) of hepatic lobules in the rabbit as an aid to hepatic surgery [98]. Some other investigations concerned experimental animal tumors. No temperatures were recorded.

In 1979 Tabuse reported a microwave tissue coagulator [99]. Temperatures were not recorded.

In 1981 a few clinical cases were described by Tabuse and Katsumi [100]. Recently, a microwave coagulator has been developed in Sweden (Risman, P.O., personal communication 1981), intended for use in open surgery.

We conclude that, in the available literature no investigations are reported, which concern the combined invasive and surgical application of microwaves.

Additional literature [101]

Rationale for investigating invasive surgical application of microwaves in the management of urinary bladder carcinoma

Bladder cancer is a common malignancy (Sweden 1976:1400 cases) with increasing incidence (males +3.7%; females +2.1% per year). Superficial cancer is usually treated by transurethral electroresection. Recurrences are frequent, but can in general be treated transurethrally. In some cases, recurring tumors are more aggressive, and infiltrate the muscular layer of the bladder wall. The Nd-YAG-laser can be used as an alternative to electroresection in small, superficial tumors, and as an adjuvant in larger tumors.

Primarily or secondarily infiltrating carcinoma carries a poor prognosis. Within one year, about 40% of these patients have died from their cancer. These tumors cannot be eradicated by conservative means. The most efficient treatment is combined radiotherapy and radical cystectomy. Removable tumors in inoperable patients are usually treated by intendingly curative radiotherapy. When the tumor is in an inoperable stage, radiotherapy is usually inefficient. In case of severe local symptoms, these patients can be offered palliative radiotherapy.

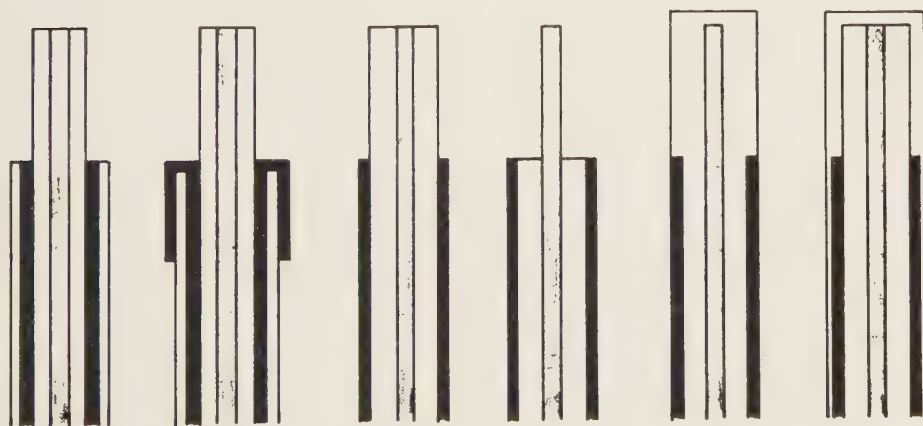
Superficially widespread cancer, which is seen in few patients, can be treated with cystectomy alone. Cytotoxic drugs are of little value in bladder carcinoma.

The treatment of bladder cancer puts high demands on the patient and on medical care. The combined therapy is mutilating, leads to a certain disability, and is associated with a considerable frequency of urinary tract and bowel complications, even if the malignancy has been cured. Inoperable patients and those with residual or recurrent carcinoma often have severe local symptoms, such as pains and bleeding. At present, there is no qualifying therapy for these patients. We therefore find that there is a need to develop improved techniques for bladder cancer therapy. Primarily, a therapy for infiltrating, inoperable cancers with bleeding is required. Large in depth heating may carry a future possibility to avoid mutilating surgery in some patients.

Construction of an invasive microwave antenna. Temperature measurements and determination of heating patterns

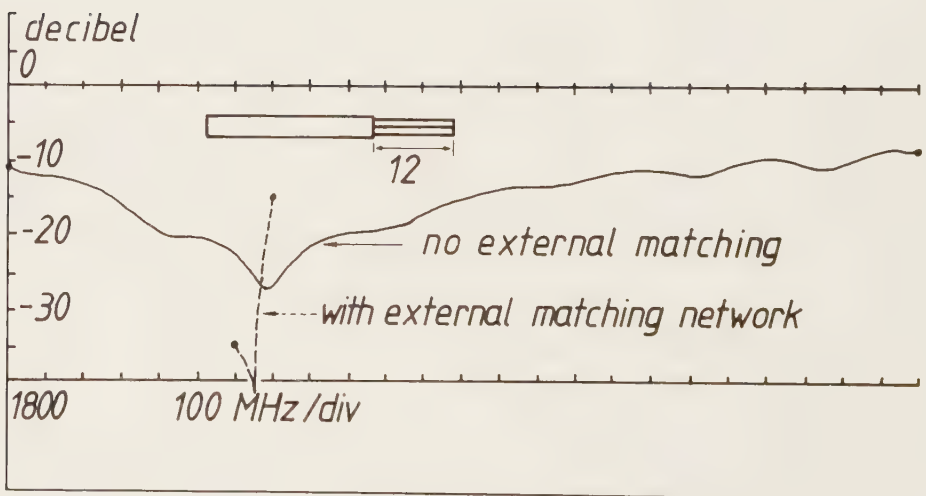
A simple and inexpensive power source was found in the 2 450 MHz magnetron. The next investigation concerned the physical dimension of the antenna. In order to allow its insertion through a standard cystoscope, the outer diameter must not exceed 2.5 mm. A semirigid coaxial cable with extended inner conductor was selected. The prototype was constructed from a copper semirigid cable (Precision Tube, BA 500085), with an outer diameter (OD) of the outer conductor of 2.16 mm, and an OD of the single strand inner conductor of 0.5 mm. This cable is uninsulated. A review of the literature at that time offered no help with exception of the standard monograph [102] which gave some theoretical solutions for antennas in lossy media. Since only a near field was expected to be useful and no information on the optimization of a near field was available, the antenna was designed experimentally. Because of our original intention to use a few minutes for heating to 55°C, the power demand should be in the order of 15 W. This produced an unacceptable heating of the tested flexible cables, and therefore this approach was postponed.

In order to find the proper length of extension of the inner conductor, for obtaining optimum matching conditions, in a lossy medium, corresponding to muscle tissue, a phantom material was used. The material (Whamo Superstuff) will be referred to as "Superstuff". The dielectric and thermal properties are very close to those of muscle tissue but have somewhat different temperature dependences. The initial designs of the antenna were the simplest possible – the outer conductor was removed for varying lengths. Later, other antenna designs were tested (Fig. 49). Using an automatic net work analyser (HP-

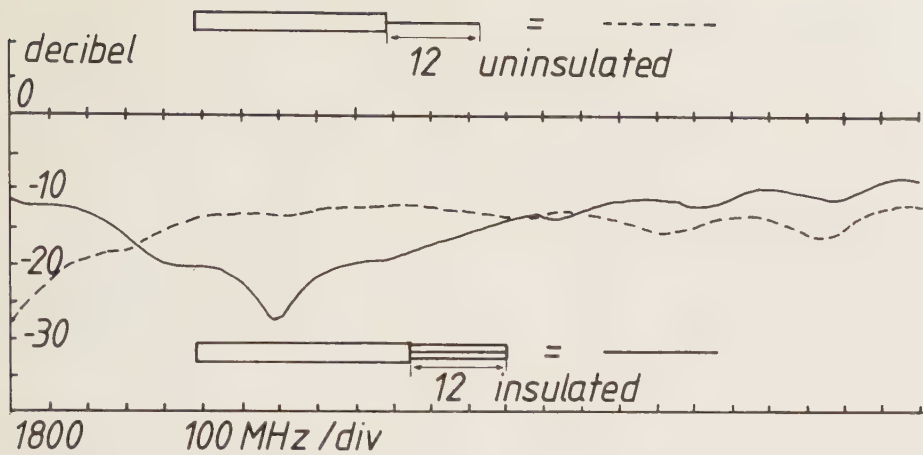


- ▣ = INNER CONDUCTOR
- = OUTER CONDUCTOR
- = DIELECTRIC

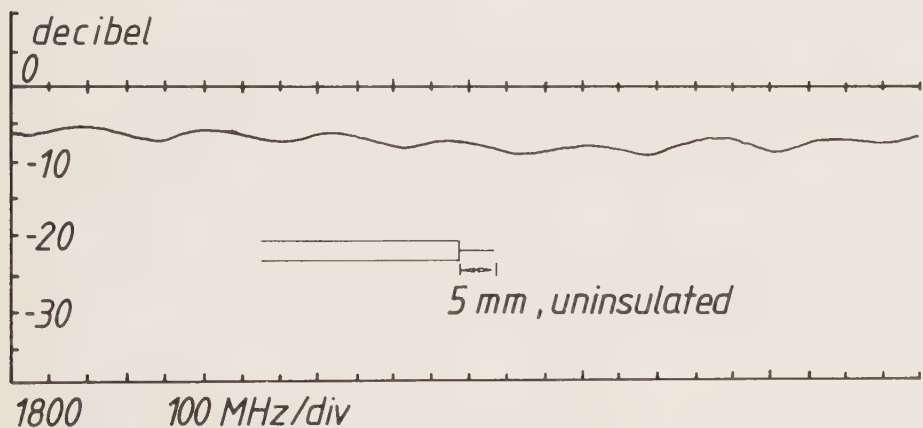
8410B), the frequency was swept from 1 800 to 4 200 MHz, and the reflected power recorded with the antenna inserted into Superstuff. This was performed without an external matching network. A narrow sweep (2 400–2 500 MHz) was measured after external matching with two double-stub tuners. This improved the matching considerably, but decreased the bandwidth (Fig. 50). We considered a reflected power of -10 to -20 dB acceptable for our application. We also found it important to maintain a broad bandwidth, since the dielectric properties of tissue might change during heating. With external matching, this could produce variations in the radiated power, which at high power would be disastrous. It was found that this antenna design at 2 450 MHz produced no more than -20 dB reflected power if the inner conductor was extended 12 mm (Fig. 50). The design with a bare inner conductor proved to be unsuitable (Figs. 51–52). It was also found that the matching of the flexible cable improved significantly when a flange was added (Fig. 53). Heating patterns were determined by different techniques. The gelatine technique proved impractical, since the temperature of interest was about 60°C , and gelatine has a considerably lower melting point. Egg white proved to be practical, since it indicates 58°C by denaturing. Convection movements in the liquid distort the coagulum but this error is minimized if the antenna is horizontal and the coagulum is viewed in the vertical plane. Some results are shown in Fig. 54.



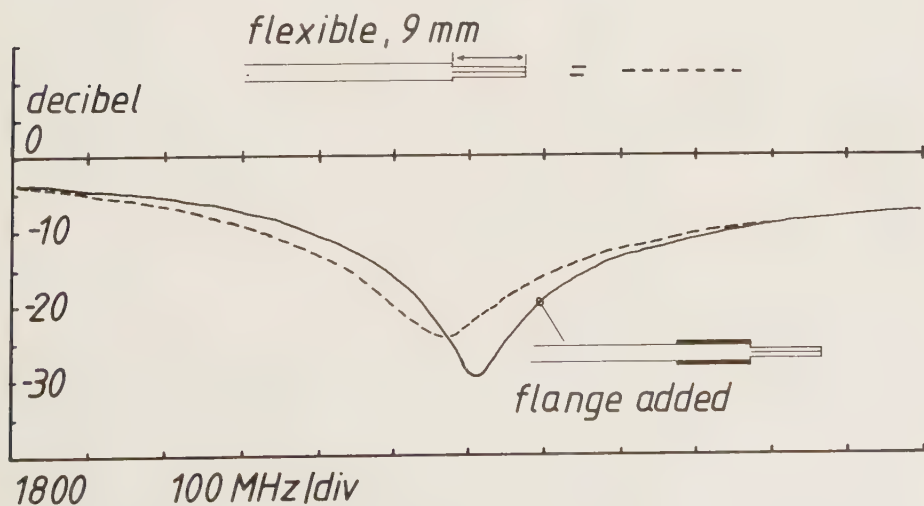
50. Decrease in bandwidth with external matching network. Reflected power versus frequency.



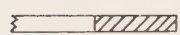
51. Insulated vs uninsulated inner conductor.

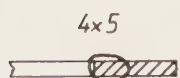


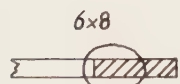
52. 5 mm uninsulated inner conductor.

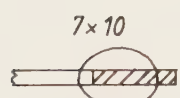


53. Flexible cable antenna with and without a flange.

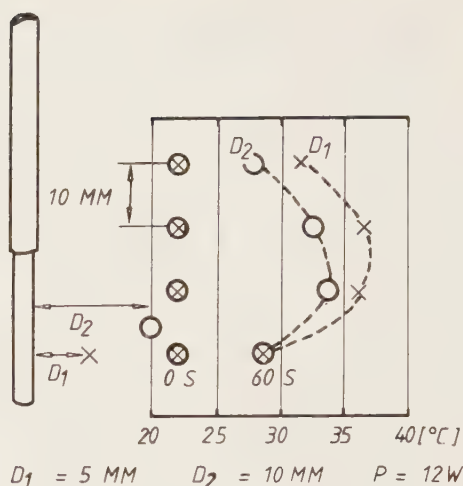
 0 s; 0 W

4x5
 2 min; 15 W

6x8
 3.5 min

7x10
 7 min

54. Heating patterns in egg white.

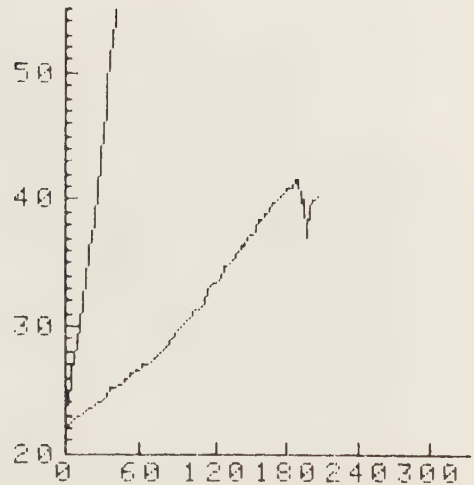
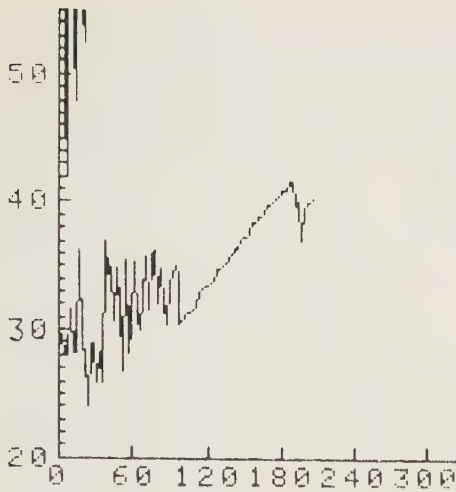


55. Temperature profile in Superstuff. 60 s power on.

Heating patterns were also determined by measuring the temperature in Superstuff. The time/power-temperature relationship is of major interest in our application, whereas steady state conditions are less interesting. Some results from 60 s heating are shown in Fig. 55. The axially long ellipsoidal shapes produced agree with expected temperature distributions, and with measurements reported by others [83]. Heating patterns by coagulation (58°C) in fresh raw meat were also studied. It should be noted that a certain thermal conduction occurs with heating times of more than about 60 s. In these experiments, the large penetration depth makes conduction important after 90–120 s of heating.

In our temperature measurements we used miniature thermocouples. There are two risks for obtaining erroneous readings, namely induced current in the leads, and field perturbation [103]. These errors are minimized by reducing the wire diameters. Our thermocouples were copper-constantan wires of 0.076 mm diameter, the junction weldings performed by laser technique, with a width less than 0.15 mm. Provided that the wires were placed perpendicularly to the generated E-field, the variation of temperature, when microwave power was switched off and on, was less than $\pm 0.5^\circ\text{C}$. With the thermocouple close to the antenna, a significant disturbance will occur if the wires are longitudinal to the antenna (Fig. 56). It has been claimed that field perturbations caused by thermocouples might generate hot spots. We investigated this by heating egg white at high power, with thermocouples placed close to the antenna. Any hot spots over 58°C would have been made visible by coagulation, but this did not occur.

Another error is introduced by the data acquisition system, compared to the use of tables for converting thermo-voltage into temperature. This relation was namely approximated with a 5th order polynoma and a computer was used for the conversion. This error is less than $\pm 0.2^\circ\text{C}$ in the temperature region of interest. A more serious error is due to inexact positioning of the thermocouple, which of course applies to any invasive measuring technique. With the temperature gradients encountered in our application, a positioning error of 0.5 mm may in the worst case produce an error of 5–10°C, depending on the distance from the antenna.



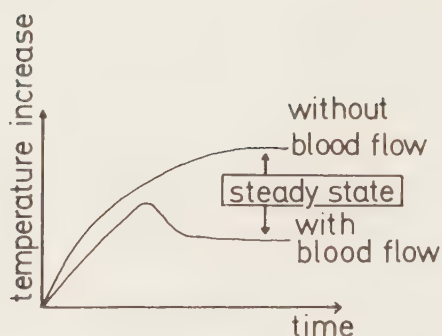
56. Disturbed measurement of temperature with thermocouple wires longitudinal to antenna (a). In (b) they are placed perpendicularly under otherwise identical conditions.

Other possible temperature probes are thermistors, which in combination with highly resistive carbon film leads are quite accurate. Several types of non-perturbing fiber optic sensors have been devised. These generally offer only one (expensive) point of measurement. Since several years research goes on in order to develop non-invasive temperature monitoring techniques, such as ultrasonic principles [104–107] or microwave-millimeterwave radiometry [108–110]. There are obviously a few obstacles to be overcome before these techniques are clinically useful.

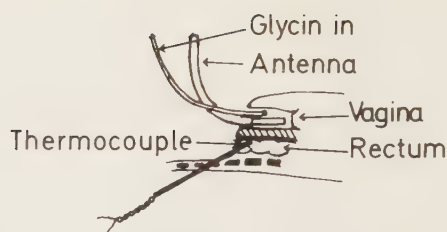
Heating patterns can be determined in different ways. For our purpose, the temperature distribution vs. time relationships are of most interest, which makes the generally applied techniques less useful. This has justified the use of only two measuring points during some of our investigations. Since we will not achieve steady state conditions in surgical applications, the mapping of temperature profiles by a single probe, moved about in the heated volume, will not yield the information needed. Most attractive would be to obtain a perfect map of the E field, which in principle could be done using a Schottky crystal diode detector. This is, however, for small heated volumes not accurate enough in the region of major interest.

Preliminary in vivo investigations in the white rat

For these investigations the microwave system consisted of a magnetron producing a rather high peak power, but a lower average power. Generated, incident and reflected power were measured using dual directional couplers and power meters (HP 432 A; Marconi 6555). The radiated power was recorded on a desktop computer (HP 85), and the losses in the transmission line were compensated for. The thermocouples previously described were used, with an ice-water cold reference. Temperatures were also recorded on the computer and plots were made with a HP 9872 B plotter. During the experiments temperature was displayed on the computer.

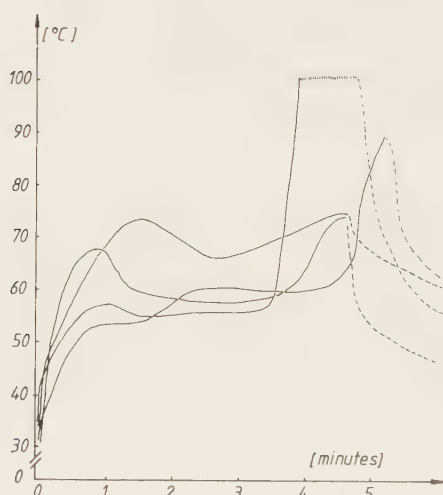


57. Classical temperature vs time plot in e.g. hyperthermia.



58. Experimental arrangement. Irradiation intravaginally with temperature recording in the anterior rectal wall. Distance to antenna ~ 3–4 mm.

As previously described [77], heating produces an initial increase of blood flow, with a maximum about 45°C . This results in temperature vs. time plots similar to the curves in Fig. 57. The stimulation of local perfusion is interesting, since it has been shown that heating of peripheral nerves produces excitation at 45°C [111]. Initial experiments with local heating of the kidney, in the anesthetized white rat, showed a similar graph which, however, was interrupted by a sharp increase of temperature after several minutes of steady state. To find out if this was reproducible, a small series of exposures in the female white rat was performed. The experimental arrangement is shown in Fig. 58. The recorded temperature vs. time plots are shown in Fig. 59. The results are



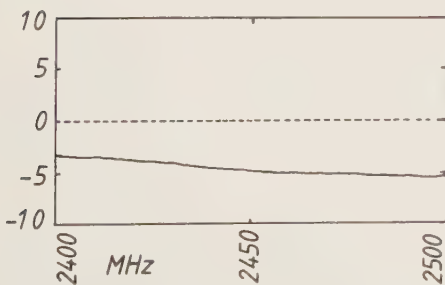
59. Temperature vs time in the experiment described in Fig. 58.

difficult to explain otherwise than in terms of vascular occlusion. This fact initiated further studies of microvascular changes in thermal surgery, which are not yet completed. The results do, however, point out that cooling is not impaired until arteries a few generations before the capillary network have been occluded (Fig. 60). The term "thermally significant vessels" has been used to describe this size of arteries [112]. In the cited analysis it was also concluded that these vessels are largely responsible for cooling, when the heat balance in a smaller tissue volume is regarded.

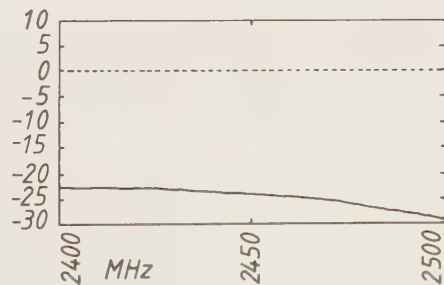


60. Microangiography in murine kidney. Area microwave heated to about 55°C for a few minutes. Main vessel (150 μ m diam.) only partly filled. Several occluded branches (50–75 μ m diam.) are seen.

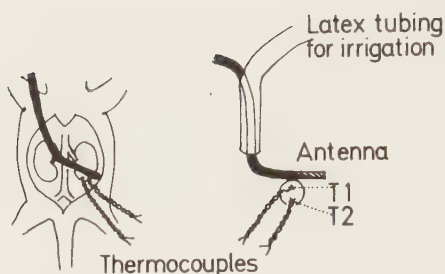
Our first experiments were performed with the uninsulated antenna. Intense heating caused by large conduction currents occurred close to the antenna. The function was improved by insulating the distal part of the outer conductor and the extended inner conductor. This made an extension of the inner conductor to 18 mm necessary for optimum matching. The reflected power of this antenna in air and muscle phantom is shown in Fig. 61. The antenna does not radiate in air, it is well matched in tissue, and has an acceptable bandwidth. The heating with this antenna proved to require higher power than with the previous design.



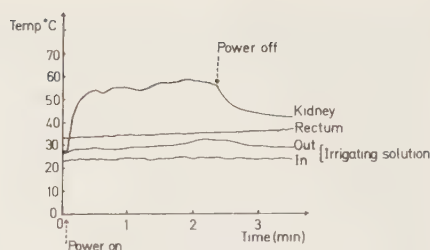
61. a) Reflected power in air for the insulated antenna.



b) Reflected power in Superstuff for the insulated antenna.



62. Experimental arrangement for standardized irradiation in murine kidney.



63. Recorded temperatures in experiment described in Fig. 62.

Next, a series of standardized radiations in the murine kidney was performed. White male rats (200–250 g) were anesthetized by ether induction and intraperitoneal chloral hydrate maintenance. The kidneys were exposed through a midline abdominal incision. The abdominal walls were lifted to enable the retention of irrigating solution (Glycin^R 22 mg/ml, Travenol, Baxter Labs.), which was supplied at room temperature and at a constant flow rate. The intraabdominal volume was kept at a constant level, excess fluid being removed by suction. During irradiation the fluid temperature was constant. At 30 W radiated power applied to the kidneys as shown in Fig. 62, the central kidney temperature was fairly reproducible at $56 \pm 2^\circ\text{C}$ (Fig. 63). 28 kidneys in 14 animals were irradiated 15, 30, 45, 60 or 90 seconds, the irradiated power being 29 ± 1 W. The animals were sacrificed at different time intervals: the 15–45 s animals 72 h after exposure, the 60–90 s group at 24 h intervals up to 144 h after exposure. The removed kidneys were transected, a specimen chosen at the plane of deepest visible injury. The specimens were immersion fixed in 10% formaline, paraffin blocks prepared and 4 μm sections stained by the Htx-Eosin technique. *Macroscopically* very small coagulative changes were seen on the kidney surface. At longitudinal transection a usually hemorrhagic coagulation zone was seen, except in the kidneys irradiated for 15–45 s, which appeared normal.

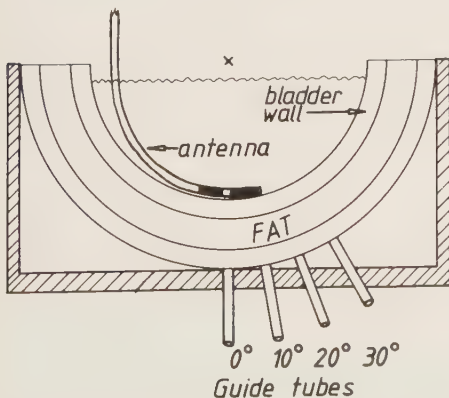
Histologically no pathology was seen in the 15–45 s specimens. Within 24 h there was hardly any recognizable damage seen in the kidneys irradiated for 60–90 s. No significant difference in necrosis development was noted when 48 and 72 h specimens were compared. The qualitative findings differed from other thermal necroses only by the remarkably small inflammatory response at the border of the necrosis.

The size of the lesions was fairly reproducible. The shape was cylindrical, with diameters of 3–4 mm in the 60 s samples and 4–5 mm in the 90 s samples. The depth was 5–6 mm in all samples. At the surface there was usually a 0.5–1 mm zone of intact parenchyma. This was first suspected to be explained by the fact that sections had been taken at some distance from the point of antenna/kidney surface contact. This, however, was not the case. The remaining explanation was that the irrigating solution produced efficient surface cooling of the kidney. This series indicates that a standardized procedure can be used. One

uncertain fact, however, is the amount of radiated energy absorbed by the tissue under different geometrical conditions. Since these experiments fairly well simulated the endoscopic situation, a too low radiated power and too efficient surface cooling, can be expected to preserve superficial tumor cells. Even if a deep vascular occlusion is produced, the nature of bladder vascularization is likely to provide collateral perfusion to the superficial region [113]. An efficient heating at the surface can be accomplished by increasing the temperature of the irrigating solution to 37°C and/or by increasing the radiated power.

Investigation of temperature distribution in a bladder phantom heated by the microwave antenna

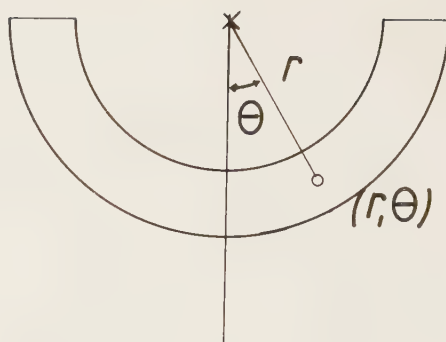
A dielectrically multilayered bladder phantom was built in a specially constructed frame. The muscular wall is normally at least 3–4 mm. In the phantom this was simulated by a 5 mm layer of Superstuff. Outside this a 10 mm layer of porcine crude lard was placed to simulate perivesical fat. Normally this layer is 4–5 mm, with the peritoneum, where applicable, adding an additional 1–3 mm distance to adjacent intestinal loops. The lossy tissues surrounding the extravescical fat were simulated by another layer of Superstuff. The frame was provided with angulated guide tubes for correct positioning of thermocouples (Fig. 64). It was built as a half-sphere. Three thermocouples were mounted on a small glass tube (Fig. 65). The positions of the thermocouples were thus defined by the angle and the distance to the center of symmetry (Fig. 66). The phantom was filled by Glycin® solution which was kept at constant temperature by continuous exchange. The antenna was positioned with its driving point (see next section) in the line corresponding to the angle $\theta = 0$, and irradiations performed with a radiated power of 13–14 W for up to 15 minutes. Temperatures were recorded for different angles and distances, r , to the phantom center. The radiated power was low, and in vivo, convection losses are expected to alter the temperature distribution significantly. The results are shown in Figs. 67–70, with marking of the temperature range corresponding to coagulation (58–63°C), if the same temperature increase had started at 37°C instead of room temperature. At an angle $\theta = 30^\circ$ no necrosis would result in the bladder wall even at 15 min of this power. At exposure times of 3–4 minutes transmural necrosis would result with $\theta \rightarrow 20^\circ$. In vivo convection losses reduce the angle within which necrosis results. With increased power, a possible application mode is suggested.



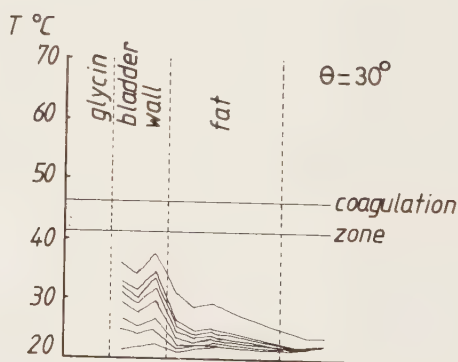
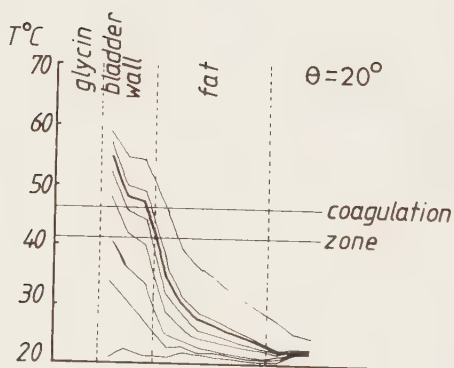
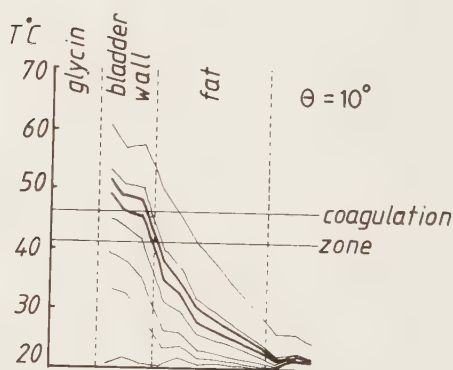
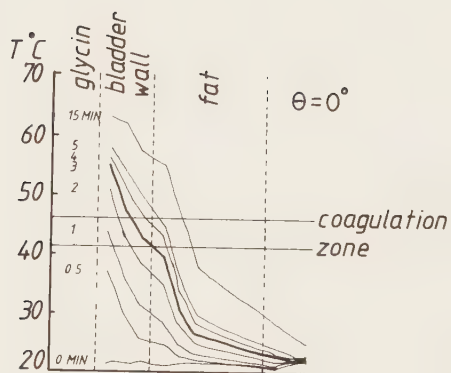
64. Bladder phantom.



65. Thermocouple mount.



66. Thermocouple position determinants in bladder phantom.



67-70. Temperature recordings in bladder phantom at different angles relative to the antenna. Spatial distribution. Temperature

graphs corresponding to transmurial necrosis marked with bold lines.

Theoretical considerations of antenna efficiency

It has now been proved that the field of this antenna can be used for controlled surgical applications. Therefore it could be valuable with some theoretical description of its near field. Maxwell's equations for the radiation field from a half wave dipole in a lossy dielectric cannot be used in the region close to the antenna, i.e. the near field. As a useful approximation the antenna can be regarded as biconical (Fig. 71). The complete solution to the E and H fields for this case consists of infinite series. The E field consists of an E_r and an E_θ component (compare Fig. 72, spherical coordinates). The only component of the H-field is H_ϕ . If we suppose that the cones are of infinite length, or that the medium surrounding the antenna has high dielectric losses, the relation between the generated power density q , the cone angle, θ_0 , the angle, θ , with the axis of the cones and the distance r to the origin is

$$q = \frac{\alpha P_0}{4\pi \ln \left(\cot \frac{\theta_0}{2} \right)} (r^2 \cdot \sin^2 \theta)^{-1} \cdot e^{-2\alpha r} \quad \text{where}$$

α = the attenuation coefficient

P_0 = the total radiated power.

This expression was used to compute the temperature distribution in the bladder phantom experiment. The experimental and the computed (using a finite element method) distributions showed a rather poor correspondence. In spite of this, we believe that this equation describes the near field better than the temperature proportional to deposited energy suggested by Taylor [90], i.e. $r^{-6} \exp \sim [2k_0(\epsilon')^{1/2}]$. In a more recent model we regard the antenna cones as being surrounded by two different dielectrics 1 and 2 (Fig. 73). In this case only the fundamental mode $E = (0, E_\theta, 0)$ and $H = (0, 0, H_\phi)$ is excited. The region $r < r_0$ is an input region, where $r_0 \ll 1$. The field within $r < r_0$ is disregarded of, and the power output is assumed to take place at $r = r_0$.

The medium 1 has $\epsilon_1 = \epsilon_0 (48 - j16)$. In agreement with the first model we assume that spheric waves are propagated, but assume that a standing wave occurs in medium 1 due to reflection at $r = 1$. Thus only waves in the positive r direction are propagated in medium 2. The boundary conditions at $r = 1$ are chosen to give total absorption of the power in the medium 1. This is comparable to a transmission line terminated by an impedance $z = \infty$.

The resulting expression for induced power density is now

$$q(r, \theta) = \frac{P_0 \alpha}{4\pi \ln \left(\cot \frac{\theta_0}{2} \right)} \frac{e^{2\alpha(r_0 - r)}}{r^2 \sin^2 \theta} \{ 1 + e^{-4\alpha(1-r)} + 2e^{-2\alpha(1-r)} \cos 2\beta(1-r) \}$$

$r_0 \leq r \leq 1$ where

β = phase shift per unit length.

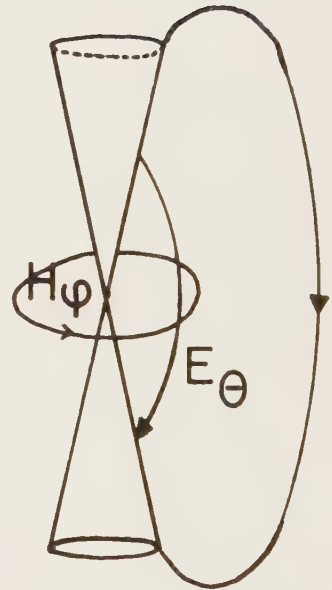
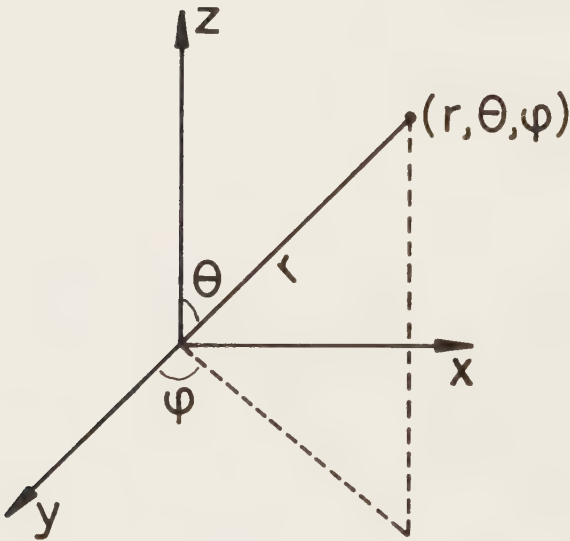
[Roos, D. Internal report 1982. Chalmers University of Technology, Gothenburg, Sweden.]

The validity of this equation has to be investigated.

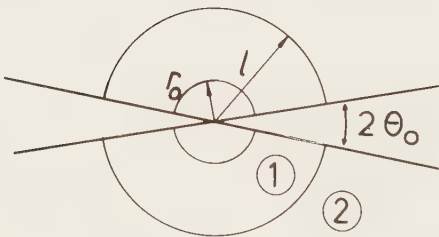


Driving point

71. Theoretical model of the biconical dipole antenna.



72. Field configuration at biconical antenna. Spherical coordinate system.

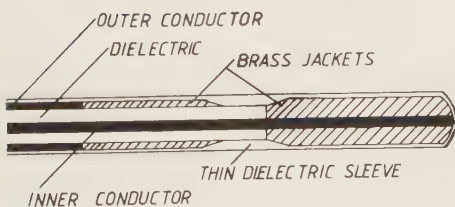


73. Conical antenna surrounded by two different media.

A 2 450 MHz generator with continuously adjustable power and feedback temperature control. Improvement of the antenna.

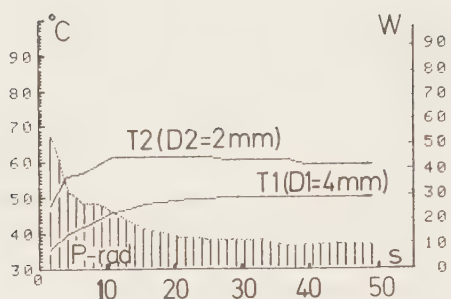
A 700 W magnetron originally designed for operation with a full wave rectified high voltage was selected. A SCR pair was introduced in the mains supply to the high voltage transformer and rectifier bridge. The thyristor pair was operated in counter phase of the 50 Hz ac, resulting in pulses of mains ac at 100 Hz with variable width. The gating of the SCR:s was controlled either by the manual setting of a dc voltage, or by a temperature dependent voltage supplied by a desktop computer (HP 85), programmed to maintain a constant temperature in a master thermocouple. Microwave energy was extracted from the magnetron rectangular waveguide by a probe. For protection of the magnetron a circulator and termination were connected to the waveguide, and the power was coupled to a N-connector on the front panel of the unit. The power was transferred to the surgical antenna by a flexible coaxial cable (RG 400/U, OD = 5.1 mm, HABIA AB, Sweden) with the connection to the antenna made by BNC-connectors. The microwave output of the generator is pulsed (100 Hz) high peak power (appr. 700 W) with the average power adjustable 0–100 W. This may produce somewhat different heating patterns than a CW output. Modifications, including a built-in dual directional coupler with crystal detectors, display for preset radiated power and a timing circuit are under way. When computer control is not used, a footswitch activates the magnetron.

The antenna was modified to obtain more uniformity between the inner conductor and the outer conductor parts. Several flexible cables have been used (RG 178 B/U, RG 317/U, HABIA AB, Sweden), with OD ranging from 1.9 to 2.6 mm. These are capable of transmitting the required power without excessive heating. The antenna and its transmission line have been encased in a thin waterproof dielectric sleeve, fulfilling also mechanical strength and hygiene demands. The design of the antenna is shown in Fig. 74.

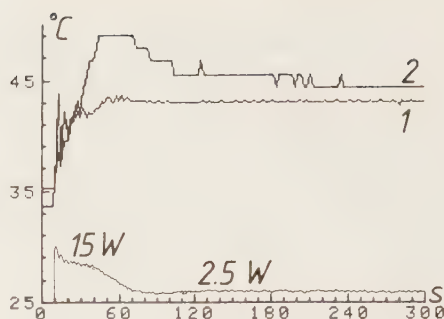


74. Present antenna design.

The computer program intended for use in hyperthermia produced a step-wise adjustment of the thyristor gating guided by the actual temperature and the increment in the preceding temperature readings. At testing in vitro and in vivo it performed well, with an accuracy better than $\pm 0.5^\circ\text{C}$ (Figs. 75–76).



75. Computer program for temperature regulation used in Superstuff. Preset temperature at $T_1 = 50^\circ\text{C}$.



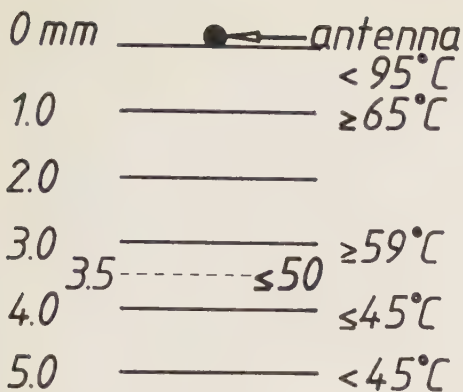
76. Computer regulator used in vivo. Murine liver. Setting 43°C . (1) = master thermocouple, 10 mm from antenna. (2) is 5 mm from antenna, and shows initially disturbed temperature. Maximum temperature in (2) 49°C .

Investigations of useful heating depths in phantom material and the white rat

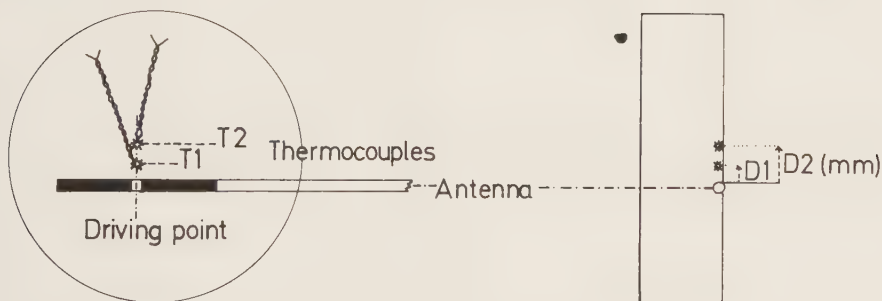
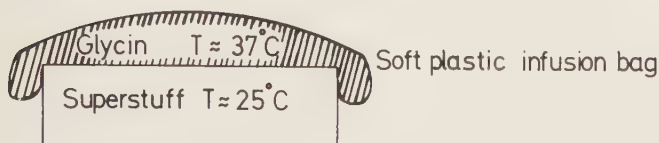
In these experiments, performed in Superstuff, the conditions were standardized as far as possible. The heat losses were reduced by using a surrounding medium at a temperature of 37°C . Since the Superstuff lost too much water at this temperature, it was used at room temperature, about 25°C . The questions to be answered were 1) Is it possible to limit the tissue damage to a depth of 3 mm (thickness of normal bladder wall)? The expected temperature profile is shown in Fig. 77. 2) How deep is it possible to heat with maintained control? This means that heating should not occur by thermal conduction, but be primarily induced by the electromagnetic field. The upper limit, Δt , for the heating time must, therefore, be of the order of $(\Delta x)^2/D$, where Δx is the penetration depth and D is the thermal diffusivity. An estimation of Δt is 30 s.

In order to find an answer to the two questions, the arrangement shown in Fig. 78 was used. In the first series, two measuring points, T_1 and T_2 , were positioned at a distance, D_1 and D_2 respectively, from the antenna. Power, time, number of pulses, D_1 and D_2 were varied. The resulting temperature graphs are shown in Figs 79–82. It is obviously possible to limit the depth of damage to between 3 and 5 mm.

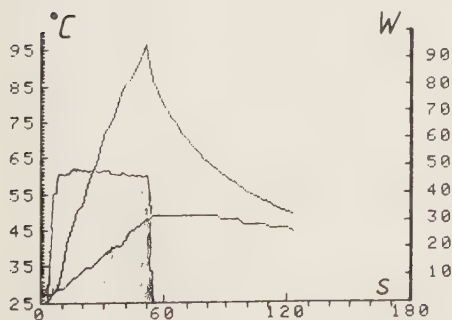
In order to investigate the second question, a single temperature measuring point was used. Heating was performed with the thermocouple at a distance of 4–8 mm. 8 recordings were taken at each distance, in order to compensate for the errors due to inaccurate thermocouple position. The variation at each distance was ± 2 – 3°C . The graphs for each distance have been averaged and collected in Fig. 83. The heating time was 30 s, with a radiated power of 50 W. The coagulation temperature is reached to a distance of 5 mm. Even at 8 mm, the temperature increase is almost instantaneous, but small ($\sim 5^\circ\text{C}$). It is possible to increase the distant temperature further, either prolonging the time of irradiation, or by increasing the power. The critical point, however, will be the subsurface temperature. The losses into the irrigating solution will decrease the surface temperature, but 1–2 mm underneath, it will exceed 100°C . This is likely to produce a crater formation, which may largely impair the mechanical stability of the bladder wall. A promising approach appears to be the use of repeated pulses. A preliminary mathematical analysis indicates that the true “penetrating depth” may be exceeded with up to 50%, without increasing the subsurface temperature too much. Further investigations of this are going on.



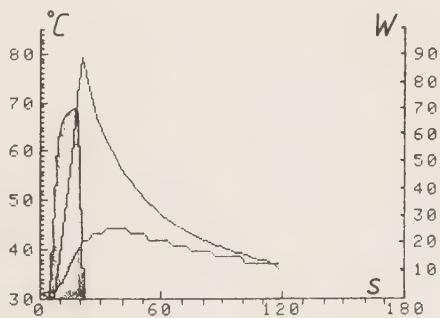
77. Temperature distribution conditions producing a 3 mm deep necrosis.



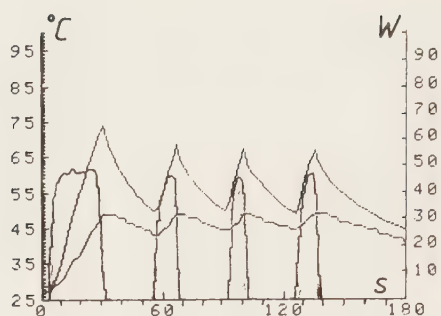
78. Experimental arrangement for investigation of in depth heating vs power-time relation in vitro.



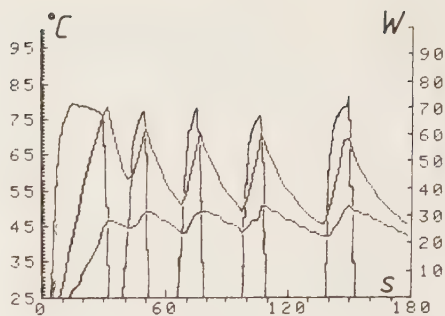
79. Superstuff. Single, wide 50 W pulse. $D_1 = 1$, $D_2 = 4$. Power off at $T_2 = 50^{\circ}\text{C}$.



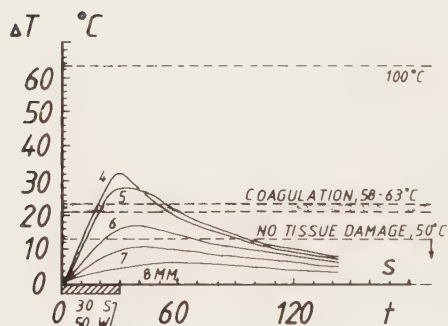
80. Superstuff. Single 70 W pulse. $D_1 = 2$, $D_2 = 5$. Power off at $T_1 = 80^{\circ}\text{C}$.



81. Superstuff. Repeated 50 W pulses. $D_1 = 1$, $D_2 = 4$. $45 \leq T_2 \leq 50$ by manual control.



82. Superstuff. Repeated 70 W pulses. $D_1 = 1$, $D_2 = 4$. $45 \leq T_2 \leq 50$ by manual control.



83. Superstuff. Averaged recordings in heating with thermocouple 4–8 mm from antenna. The temperature maximum decays with increasing distance, though increase is simultaneous with the switching on of power. The temperature maximum tends to occur later at larger distances (compare Figs 2 and 3).

This experimental evidence provides a fair background for the statement that the near field can be used for safe and controlled, yet standardized, in depth coagulation in the bladder.

The abdominal wall of the 300 g white rat with a thickness of 3 mm represents a good model of the normal urinary bladder. The animals were anesthetized by ether induction and intraperitoneal chloral hydrate maintenance. The temperature drop in anesthetized rats is usually considerable. We used some means to increase the initial temperature, to get closer to physiologic conditions in man. The rat was placed on a 1 000 ml soft plastic infusion bag, containing only 300 ml fluid, to conform to the body. The fluid was preheated to 40°C. Underneath the infusion bag was a heating blanket at 60°C. Above the rat a radiant heater was placed. To obtain antenna matching another plastic infusion bag containing preheated Glycin^(R) solution was used. In this way we could usually start irradiations at 35°C. A pilot series showed that less power than indicated by the in vitro studies was needed. With more than 50 W, the damage extended throughout the abdominal wall within a few seconds. The optimum time at 50 W seemed to be 10–20 s. The series was finished by irradiating three rats bilaterally and from the external for 10, 15 and 20 s. These rats were sacrificed after 72 hours.

Macroscopically a transmural necrosis was seen beneath the antenna driving point. It had a lateral extension of about 3 mm. A superficial lesion of a depth less than 1 mm was seen. It was ellipsoidal, the axis longitudinal to the antenna measuring about 10 mm, and the perpendicular axis 3–4 mm. Both animals irradiated for 15 and 20 showed evidence of intestinal coagulation damage, while this was not observed for an irradiation of 10 seconds.

Histologically transmural necrosis was seen in all rats. The peripheral extension depended on the application time.

Conclusions

A technique capable of deep and superficial coagulation has been developed. The prospects for controlled in depth action appear promising. Obviously, the minimum dose necessary for a 3 mm necrosis was exceeded in the experiments. Further and more extensive investigations are necessary to create a doseplanning program. The modality seems to cover the field of Nd-Yag-laser treatment, and offers possibilities for even deeper heating in infiltrating cancers. The manipulation of the antenna offers a few technical problems that have to be solved, but the used equipment has the advantage of being inexpensive compared to lasers. The microwave antenna does not radiate in air and is therefore safe for the surgeon and other medical personnel.

Lasers

The theory of stimulated emission was formulated by Albert Einstein in 1917. The maser (*microwave amplification by stimulated emission of radiation*) was developed in 1950. Schawlow and Townes presented the idea of the optical laser in 1958. In 1960 the first laser, a ruby crystal solid laser, was operated by Maiman. During the following years the Helium-Neon and carbon dioxide gas lasers, and the solid Neodymium-laser, were developed. The possibility of using lasers in surgery was realized before 1965, and the first clinical applications took place about 1970. Though laser techniques are well-established in industrial applications, their use in surgery is still young. A variety of lasers are under investigation for surgical applications, while four laser types are well-established. Three of these, namely, the CO₂ laser, the Nd-YAG laser and the Ar laser have found application in urologic surgery.

Basic laser optics

Like other electromagnetic radiation, laser beam energy is absorbed in biological tissue. The conversion of optical energy into thermal energy is the basis for surgical laser applications. The radiation is also scattered at the tissue surface and within the tissue. The degree of absorption and scattering determines the volume distribution of the absorbed power density. The laser wavelength and the optical properties of the tissue determine the ratio of absorption to scattering. The on-axis decrease in laser beam power density is described by the *extinction coefficient*, γ , which is equal to the sum of the *absorption coefficient*, α , and the *scattering coefficient*, β . The initial beam power P_o , the power $P(x)$ after the path length x in the tissue, are related by $P(x) = P_o e^{-\gamma x}$, where $\gamma = \alpha + \beta$ [m^{-1}], which is known as the law of Lambert-Beer. At a depth, $x_e = \gamma^{-1}$, $P(x_e) = P_o e^{-1}$, which means that the beam power has decreased to e^{-1} , or that $(1 - e^{-1}) P_o = 63\%$ of the beam power has been scattered and/or absorbed. x_e is termed the *mean free path* (cf microwave penetration depth). The equation, however, does not give a description of power distribution in case of significant scattering. The ratio of α to β gives an estimation of the surgical effects with different lasers, as shown in Table C. Other mathematical models have to be used to describe the power redistribution within tissue at significant scattering [114].

The laser beam has a diameter, D_m , defined by twice the distance from the beam axis, where power density has decreased to $e^{-2} = 14\%$. The beam is often not quite parallel but has a slight divergence, expressed by

$$\theta = \frac{4\lambda}{\pi D_m} \quad \text{In the fundamental transverse electromagnetic mode, TEM}_{00}, \text{ the}$$

power density distribution along D_m is Gaussian, and the focus spot size, D , is given by $D = f \theta$, where f = the focal length of the lens. The practical consequence of these equations is that a longer focal length and wavelength give a larger focal spot than a shorter focal length and wavelength.

The minimum available focal point ("spot size") with a CO₂-laser and a f=50 mm lens is about 0.1 mm, while with the Ar laser a focal spot of 25 μm can be produced. The laser beam power density in the tissue surface, can at a given power be adjusted by variation of the spot size. The optical properties of tissue change when the temperature increases. This has special implications for the Nd-YAG laser. The conditions specific to transurethral Nd-YAG laser surgery are discussed in e.g. [115].

A model for mathematical analysis of the in depth tissue damage with the CO₂ laser.

The CO₂ laser radiation is generally assumed to be absorbed practically in the tissue surface, i.e. at the depth ≈ 0 . The radiation does however, have a certain depth of penetration. The absorption coefficient for tissue is assumed to be in the order of $\alpha = 2 \times 10^4 \text{ [m}^{-1}\text{]}$. Since scattering effects can be neglected, the Lambert-Beer law is applicable. Thus at the depth d,

$$\begin{cases} P_{\text{abs}}(d) = P_{\text{in}}(1 - e^{-\alpha d}) \\ P_{\text{transm.}}(d) = P_{\text{in}} e^{-\alpha d} \end{cases}, \text{ where}$$

P_{abs} = the absorbed power

$P_{\text{transm.}}$ = the transmitted power

P_{in} = the incident power at $d = 0$ (tissue surface)

At the depth $d = 3 \times \frac{1}{\alpha}$, $P_{\text{abs}} = P_{\text{in}} \times 0.95$. Most of the power thus has been absorbed at this depth, which is 150 μm. The transmitted power may however be sufficient to produce tissue damage to deeper layers. The applied model consists of 6 layers, each having a depth of a mean free path length, 50 μm. The spot size is chosen to 0.5 mm. The incident power has been chosen to produce a minimum of tissue removal, and to produce maximum heating without vaporization of water in the 4th layer. With a higher incident power, more tissue will be removed, but the conditions at the boundary region are likely to be similar. In Fig. 84 an exploded view of the model is shown. Some of the values used for the calculations had to be assumed, since tabulated values for all the substances could not be found. These values are included mainly in the terms having less significance in production of errors. The exposure time, 50 ms, was chosen to exclude significant thermal conduction phenomena. The loss of thermal energy from the extremely hot tissue surface by blackbody radiation was estimated by Stefan-Boltzmann's law, $Q_r = \sigma T^4$, and found to be in the order of 10^{-6} W , and thus negligible. The successive events in producing a temperature expected to disintegrate tissue, i.e. to remove it, are assumed to be:

1. Heating of tissue to 100°C
 $\Delta T = 100 - 37 = 63\text{K}/C_p = 3 \times 10^3/\varrho = 1.07 \times 10^3$
2. Vaporization of water in the tissue
 $C_{\text{evap}} = 2.253 \times 10^6/\varrho = 10^3$
3. Heating of dehydrated tissue to carbonization, 200°C
 $\Delta T = 200 - 100 = 100\text{K}/C_p = 0.75 \times 10^3/\varrho = 0.07 \times 10^3$
4. Heating of carbonized tissue to disintegration, 450°C
 $\Delta T = 450 - 200 = 250\text{K}/C_p = 0.5 \times 10^3/\phi = 0.5 \times 10^3$

Assumed fractions of

$$\begin{cases} \text{Water to normal tissue} = 0.7:1 \\ \text{Dry tissue to normal tissue} = 0.3:1 \\ \text{Carbonized tissue to normal tissue} = 0.2:1 \end{cases}$$

The absorbed energy in each volume element is averaged over the entire volume to calculate the resulting temperature. This introduces a small error, mainly concerning the temperature gradient over the elements $d_3 \rightarrow d_4$. To define the conditions, the element d_4 is first regarded. This element, as seen in Fig. 84 absorbs $0.0315 P_{\text{in}}$. To find the P_{in} corresponding to tissue heating to 100°C, we obtain the following equation:

$$\underbrace{3 \times 10^3}_{C_p} \times \underbrace{1.07 \times 10^3}_{\varrho} \times \underbrace{5 \times 10^{-5}}_{\text{depth}} \times \underbrace{6.25 \times \pi \times 10^{-8}}_{\pi r^2 = \text{area}} \times \underbrace{(100 - 37)}_{\Delta T} = \underbrace{0.0315}_{\text{relative absorp-}} \times \underbrace{P_{\text{in}}}_{\text{incident power at}} \times \underbrace{50 \times 10^{-3}}_{\text{exposure time tissue surface}}$$

which gives $P_{\text{in}} = 1.26 \quad [\text{W}]$

Now, we regard the energy requirement for heating elements d_1 through d_3 according to 1)–4).

We find

$$W_{d_{1-3}} = \sum_{\substack{d_1 \\ d_3 \\ d_{1-4)}} W_{1-4)} = 5.2819 \times 10^{-2} \quad [\text{J}]$$

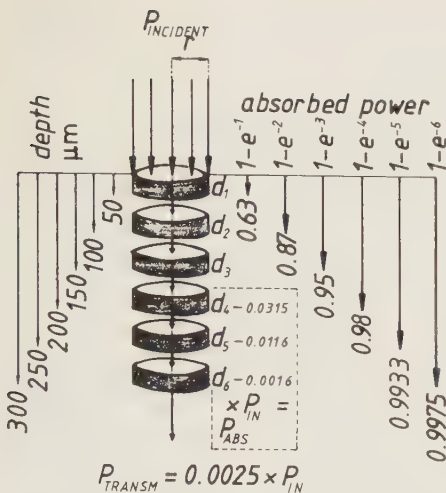
P_{abs} in elements d_{1-3} is $0.95 \times P_{\text{in}} = 1.197 \quad [\text{W}]$

The deposited energy is $W_{\text{dep}} = 1.197 \times 50 \times 10^{-3} = 5.9850 \times 10^{-2} [\text{J}]$ which just about covers the assessed requirement $= 5.2819 \times 10^{-2} [\text{J}]$. Now we will find out what the temperature increase in the element d_5 is. We find that $\Delta T \approx 23 \text{ K}$, which gives a resulting temperature of 60°C.

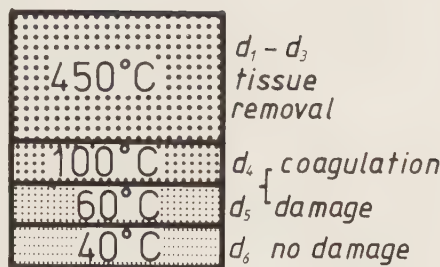
In d_6 we analogically find that $\Delta T \approx 3\text{K}$, and the resulting temperature 40°C. The temperature profile is demonstrated in Fig. 85.

The integration of temperature over d_{1-3} makes it probable that a temperature gradient between 450°C and 100°C in d_4 exists, producing a zone of carbonization, without immediate tissue removal. Equally, a gradient between 100°C and 60°C at d_{4-5} can be expected. As vessel occlusion can be expected to occur in the region superficial to the level of 70°C, the avascular zone can be expected to measure about 50 μm . The depth, to which tissue is removed, can be estimated to 100–150 μm .

Literature [116–123]



84. Exploded view of mathematical 6-layered tissue model for analysis of laser induced tissue damage.



85. Computed temperature profile with a 1.26 W; 50 ms; 0.5 mm spot size CO₂ laser pulse.

Experimental investigations of the vascular damage in CO₂ laser surgery in the murine kidney

Material and method

Five white Wistar rats weighing about 250 g were anesthetized by ether induction and intraperitoneal chloral hydrate maintenance. The kidneys were exposed through a midline abdominal incision. A Sharplan 733 CO₂ laser was used. In three rats inferior pole renal resections were performed on both sides, using a minimum (~ 0.1 mm) spot size and 10 W of continuous power. The arterial supply to the kidneys was not clamped. The incision edges were retracted, to minimize exposure of individual tissue portions. In three of the resections there was no bleeding, while in the other three heavy bleeding occurred. This was arrested by gentle digital compression and irradiation with the defocused laser beam. In two rats, irradiation was performed with about 5 W power, in 50 ms, 0.5, 1 and 2 s pulses, and in one kidney 10 W and 0.5 s. The spot size was about 0.5 mm. Several lesions were produced in each kidney.

After irradiation or resection, the abdominal aorta and the inferior caval vein were mobilized and ligated just above the bifurcation. A few mm cranially, the aorta was mobilized and cannulated with a short specially prepared thinwalled teflon catheter (Viggo AB, Helsingborg, Sweden, OD 1.00 mm). Another ligature was tied upon the cannula. The aorta was then ligated above the renal arteries. The system was flushed with a few ml of normal saline. About 2 ml of a slowly curing, dyed, elastomer (MICROFIL[®], Canton Bio-Medical Products, Boulder, Col., USA) were then injected, using a moderate pressure. The animal was killed and the closed cannula left in. The contrast medium was allowed to cure for 24 h at 8°C. The kidneys were removed, and gradually dehydrated in ethylic alcohol, 48 h each at 25, 50, 75, 95 and 99.5% by weight. The kidneys were transected, dried with tissue paper and immersed

in methyl salicylate, which makes all tissue except carbon and elastomer filled vessels transparent. After about one week, 1 mm sections were prepared using razor blades, through the center of the produced lesions. The specimens were examined and photographed in a stereoscopic dissection microscope with variable magnification under incident intense light. At 5×10 times magnification, the depth of the carbonized surface layer and the avascular zone, as well as vessel diameters were measured, using a calibrated eyepiece graticule.

Results

In the resections with bleeding, the bleeding arteries measured $350\text{ }\mu\text{m}$ in diameter. They were found to be patent until $100\text{ }\mu\text{m}$ beneath the carbonized zone, where slight extravasation of the elastomer was seen. Surrounding the vessel there was an avascular zone of $850\text{ }\mu\text{m}$ depth and $1000\text{ }\mu\text{m}$ width.

In the corresponding portion of the large artery, the lumen was contracted and contrast filling irregular, indicating thermal shrinkage and thrombus formation. The carbonized zone was $500\text{--}750\text{ }\mu\text{m}$. (Fig. 86.)

In the kidneys where a clean cut had been performed, the avascular zone was only $360\text{--}480\text{ }\mu\text{m}$. Vessels down to $10\text{ }\mu\text{m}$ were patent all the way to the avascular zone. Vessels of up to $200\text{ }\mu\text{m}$ were occluded. The lumen was decreased for only some $10\text{ }\mu\text{m}$. No extravasation of contrast was seen. The carbonized zone was $60\text{--}100\text{ }\mu\text{m}$. (Fig. 87.)

In the pulse irradiated kidneys the following was found.

50 ms, 5 W, 0.5 mm: The carbonized zone measured $30\text{ }\mu\text{m}$, the avascular $30\text{ }\mu\text{m}$. Less than $100\text{ }\mu\text{m}$ of tissue had been removed. This part of the experiments corresponds to the previous mathematical analysis.

0.5 s, 10 W, 0.5 mm: The carbonized zone was $240\text{ }\mu\text{m}$, the avascular $100\text{ }\mu\text{m}$. About 1 mm of tissue had been removed.

0.5–2 s, 5 W, 0.5 mm: The carbonized zone was $60\text{ }\mu\text{m}$, the avascular $120\text{--}240\text{ }\mu\text{m}$, but did not seem to correlate to exposure time. $0.5\text{--}1\text{ mm}$ of tissue removed.

Comments

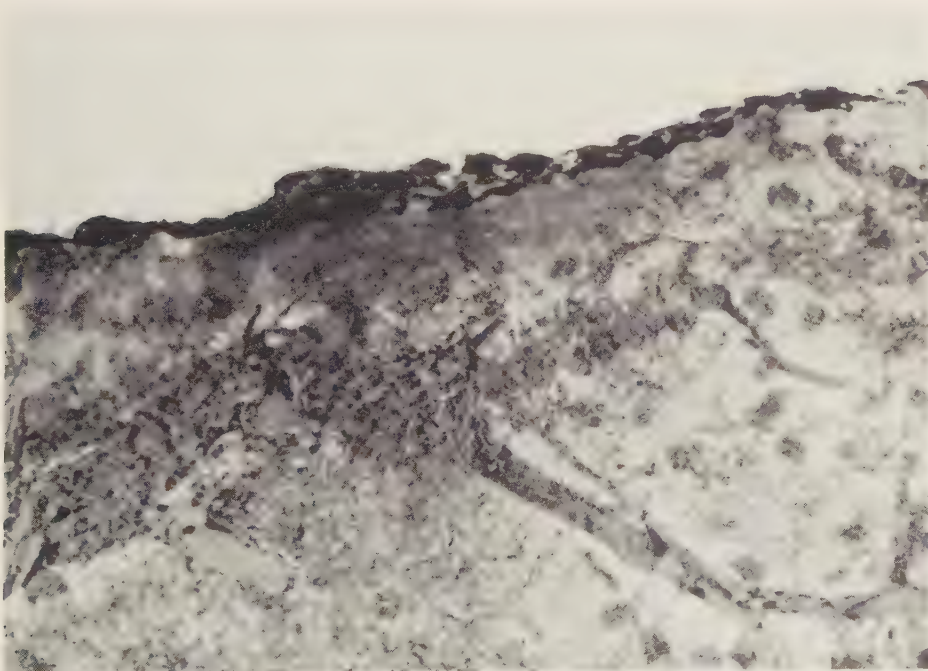
The depth of vascular damage obviously depends on the surgical technique. Particularly, prolonged attempts to produce hemostasis with a defocused beam, may produce excessive tissue damage. However, the depth of injury is less than that occurring in electrocoagulation, using non-coaptive techniques. The experiments also indicate that at least the vascular damage is reduced if short pulses are used instead of continuous power.

Discussion

The indicated efficiency of short pulse techniques being superior in limiting the tissue damage is in accordance with our clinical experience. We have removed a number of benign naevi for cosmetic reasons. Before discovering that anesthesia generally is unnecessary with pulse irradiation, we used a $1\text{--}2\text{ W}$ continuous power. When we later changed to using 50 ms pulses with a pulse power of $15\text{--}32\text{ W}$ and a small spot size, we found that epithelization was significantly more rapid, as well as scar maturation. The technique has the only disadvantage of being time-consuming.



86. Microangiography of CO₂ laser resected murine kidney. Deep vascular injury due to attempts to arrest bleeding with the defocused laser beam.



87. Microangiography of CO₂ laser resected murine kidney. Superficial vascular injury in clean cut.

How does the vascular damage correlate to tissue damage? The mathematical analysis corresponds well to the experimental studies. Maybe a deeper injury, and tissue removal, should be expected at the higher power level. That this did not occur can be partly explained by the fact that the water vapor is projected into a plume in the direction of the laser beam, and absorbs some of the incident power. This is a well-recognized problem in industrial laser applications. Besides, the control of the spot size is inaccurate. An increase of the spot size by a factor 2 produces a 4-fold increase of the energy required to perform the same surgical action. A certain amount of thermal diffusion is also responsible for the depth of the injury. The formation of a water vapor film preceding the zone of higher temperature causes a thermal insulation of the deeper structures by a phenomenon known as the Leiden-Frost effect. It can be assumed that the superficial parts of the contrast filled vessels have been exposed to temperatures in the 70–90°C range. This makes it probable that the tissue damage does not extend deeper than 50–100 μm from the boarder of the avascular zone. A complete knowledge of these facts requires further investigations.

Literature [124]

Clinical experience of a CO₂ laser in urologic surgery

A Sharplan 733 CO₂ laser has been used at our department for about 2 years. Since the unit has been subjected to transports between separate buildings twice weekly technical disturbances might have been expected. This has however, only occurred occasionally. Our clinical material is presented in Table D.

I. Treatment of condylomata acuminata with the CO₂-laser

Though a non-dramatic disease it may cause considerable discomfort to the patient and may be difficult to treat.

Material

All patients showing poor response to podophyllin applications were referred to us from the department of dermatology. A large number of the patients had been treated with more than 20 applications over a period exceeding 12 months. A total of 90 males have received laser therapy. About 80% of the patients had urethral condylomata.

Method

The patients were examined for urethral, scrotal, perineal or inguinal lesions. Treatment was performed on an outpatient basis. Initially we used local anaesthetics and continuous laser power of 1–2 W with a spot size of 0.1–2 mm to vaporize the condylomas down into healthy tissue. Later we used 50 ms pulses, at a repetition rate of 2/s, with a pulse power of 10–25 W and a spot size of 0.1–0.5 mm without anaesthesia. 2 or 6 times magnification was helpful in locating latent lesions, produced by previous podophyllin applications. Meatal condylomas were usually made accessible by simple eversion of the meatus. Those in the fossa navicularis were generally reached by the use of a nasal speculum. In some cases a short meatotomy was performed. One case with extensive lesions required a 3 cm urethrotomy. In a few cases slight bleeding disturbed the procedure. By digital compression it was possible to finish the treatment.

Results

Recurrence or remaining disease 3 months after treatment was seen in somewhat more than 5% of the patients. In some of these cases we found that the sexual partner had condylomas.

Postoperatively, no bleedings occurred. Discomfort was negligible. Epithelization was usually complete within 1–2 weeks.

Discussion

Management of condyloma often presents a difficult therapeutic problem. In patients with small glandular or preputial lesions, a few applications of podophyllin are often efficient. If these patients have recurrences, an external source of re-infection or undetected urethral disease should be suspected. Several approaches have been tried in urethral condylomas. The non-surgical are application of podophyllin, colchicin or 5-fluorouracil. Local irritation is strong with these medications, there is a risk of scarring and meatal stenosis, and they cause significant discomfort. When these patients are referred to the urologist, therapy usually consists of cryo-coagulation, electrosurgical fulguration or coagulation. Cold curettage has also been used. These methods are painful, cause considerable postoperative discomfort, and carry a risk of meatal stenosis. The therapeutic efficiency is uncertain. The Ar laser and the Nd-YAG laser have been used in treatment of condyloma.

The results obtained with the Ar-laser are probably comparable to those of the CO₂ laser. The Nd-YAG laser produces a deeper tissue damage, though at a lower temperature. The deep damage may carry a risk for stenosis. With the CO₂-laser, the temperature in the deeper parts of the condyloma is high enough to kill residual virus, which is uncertain at the temperatures produced by the Nd-YAG laser. The tissue removal with the CO₂-laser facilitates radical treatment. In the light of our experience this seems to be a superior treatment in condyloma therapy. In small condylomata it may also have advantages as a primary single session treatment modality. To investigate this compared to podophyllin, a randomized study has recently been initiated. Recently having taken a Nd-YAG laser into use, we also plan to compare CO₂-laser and Nd-YAG laser therapy.

II. Experience of the CO₂-laser in various surgical procedures

Laser surgery has been reported advantageous in patients with coagulation disorders. One patient with a moderately severe von Willebrand's hemophilia required a circumcision for secondary phimosis. We were able to perform the operation without giving blood transfusions or antihemophilic factor. There was no pre- or postoperative hemorrhage. Six patients with prostatic carcinoma were anticoagulated because of previous thrombo-embolic disease. We performed subcapsular orchiectomy in these patients, without cessation of anti-coagulation. There was no bleeding. Urethral prolapse or caruncle was treated in 6 female patients. Postoperative course was smooth, with rapid healing.

A few kidney resections were performed under clamping of the renal artery. We are not too impressed by the CO₂-laser in this application.

A 25-year-old female had persisting silent macroscopic hematuria. Bleeding was seen from the right ureter, where the IVP showed double collecting systems. We were not able to disclose the source of bleeding preoperatively, though a hemangioma was suspected. At operation, we found no bleeding from the superior pelvis, while the inferior contained blood clots. Continuous bleeding was seen from the fornices of two calyces. The corresponding papillae were evaporated with the laser, the calyces resected and the renal pelvis reconstructed. Bleeding has not recurred 12 months postoperatively, and the IVP is normal. Without the aid of the laser, resection of a major part of the kidney had been necessary.

In some patients, who had previously been subject to repeated renal surgery for staghorn calculi, the laser significantly facilitated dissection of the kidney in the scarry tissue. Nephrotomies were also performed without bleeding. One patient with a bladder carcinoma had received 60 Gy of external radiotherapy. At recurrence of the cancer, he had severe local symptoms with massive bleeding. A salvage cystectomy was performed with the laser. Bleeding during operation was less than 300 ml. The local symptoms were relieved, but unfortunately the patient died from disseminated malignancy within a few months.

A 40-year-old female had received interstitial and external radiotherapy for a cervical carcinoma, and later underwent radical hysterectomy because of recurrence. After this, she developed a vesico-vaginal fistula. This was laser excised, and the completely dry resection surface allowed precise identification of the vesical and vaginal layers. In this case, the laser facilitated the procedure.

Comments to the use of lasers in urology

The three lasers presently used in urologic surgery have quite different optical and surgical properties. Several fields of application have been tried. From our point of knowledge, I would suggest the following indications:

I. CO₂ Laser

1. Open surgery

- a) Method of choice for treatment of condylomata acuminata
- b) Removal of malignant and premalignant lesions in the external genitals
- c) Surgery in scar tissue and tissue affected by ionizing radiation
- d) Surgery in patients with coagulation disorders

2. Transurethral surgery

- a) Techniques for this application are still in the developmental stage, but may make a new modality available in the management of urethral strictures

II. Nd-YAG laser

1. Open surgery

- a) Treatment of condylomata acuminata
- b) Conservative treatment of penile carcinoma

2. Transurethral surgery

- a) Destruction of superficial bladder tumors
- b) As an adjunct to TUR in prostatic carcinoma
- c) Urethral tumors
- d) Urethral strictures
- e) Destruction of calculi???

III. Ar ion laser

1. Open surgery

- a) Treatment of condylomata
- b) Destruction of premalignant lesions of the external genitals

2. Transurethral surgery

- a) Urethral strictures
- b) Very small bladder tumors
- c) Urethral tumors

Literature [125–132]

Summary

Laser surgery requires techniques of surgery and assistance that are different from the ones we are used to. It takes time to learn these techniques and to optimize the clinical use of different lasers. Not until this has been done, is it possible to make a proper evaluation of their individual fields of application, and to define their position in our armamentarium.

Even a seemingly well-established technique, such as electrosurgery, can be further developed into expanded applications. The introduction of microwave techniques in surgery is still in a very young developmental stage.

Particularly in the light of financial development, it seems important that the different thermo-surgical modalities are developed in close collaboration between physicians and physicists. This is essential in order to define the limits and optimum range of application for each modality. This must be followed by controlled clinical trials according to carefully designed protocols. This requires the formation of inter-disciplinary research groups, devoted to the task. The clinical trials have to be performed at medical centers with access to the different modalities and established inter-disciplinary research. Such projects also require a specialized and continuous surgical supervision. Our department now has access to electrosurgery, CO₂ laser and Nd-YAG laser as well as equipment for microwave surgical applications. Inter-disciplinary research is well-established. As reported also by others (133) a proper funding of basic research, technical development and clinical evaluation is crucial in this kind of work.

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Tables

Tissue	Wavelength in tissue (mm)	Depth of penetration (mm)	ϵ'	σ
Muscle	17.6	17.0	47	2.2
Fat	52.1	112	5.5	0.2

Table A Dielectric properties of fat and muscle tissue at 2 450 MHz. The wavelength in air is 122 mm.

Frequency MHz	Depth of penetration mm
27	~ 140
433	~ 40
915	~ 30
2 450	~ 17
10 000	~ 3

Table B Penetration depth in muscle tissue for electromagnetic waves at different frequencies.

Laser	$\lambda \mu m$	αm^{-1}	βm^{-1}	α/β	mean free path mm	relative volume distribution
CO ₂	10.6	$\sim 10^4$	≈ 0	$\gg 1$	0.05	Extremely small
Ar	≈ 0.5	—	—	≈ 1	0.3	Small
Nd-YAG	1.06	~ 10	$\sim 10^3$	$\ll 1$	1.0	Large

Table C Some characteristics for different lasers in moderately vascular tissue [115].

Erythroplasia Queyrat	2
Cystectomy	2
Vesicovaginal fistula	1
Skeletal resection	1
Bladder leukoplakia	1
Cancer testis	1
Local excision of tumor	1
Benign skin tumors, removed for cosmetic reasons	25
Condylomata acuminata (♂)	90
Subcapsular orchiectomy	13
Urethral prolapse	6
Circumcision, meatotomy	10
Laparotomy	4
Kidney resection	3
Transplant nephrectomy	1
Penile carcinoma (amputation)	2
	163

Table D Clinical CO₂ laser surgical material. Dept of Urology. Lund. 1980–1982.

